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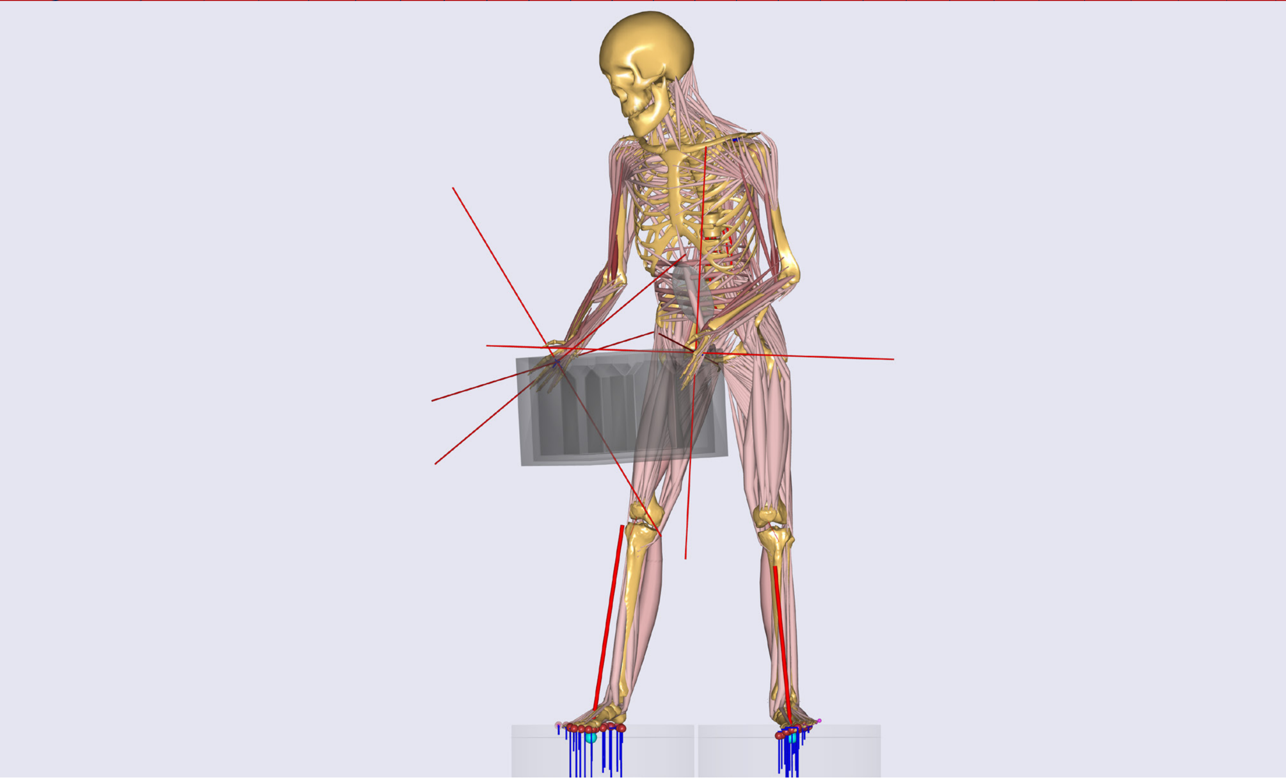

MUSCULOSKELETAL MODELLING OF MANUAL MATERIAL HANDLING IN THE SUPERMARKET SECTOR

BY
SEBASTIAN SKALS

DISSERTATION SUBMITTED 2021

AALBORG UNIVERSITY
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Publications

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ENGLISH SUMMARY

Musculoskeletal disorders are the most widespread work-related health problems resulting in tremendous societal costs. Low back pain is the most common work-related musculoskeletal disorder and constitute the number one health problem in the world measured in years lived with disability. Manual material handling (MMH) is the most consistently identified occupational risk factor contributing to the development of low back pain. Among the many industries where MMH is common, the supermarket sector has received little attention in the scientific literature despite musculoskeletal disorders being highly prevalent in the industry. This doctoral dissertation employed state-of-the-art methods for inertial-based motion capture and musculoskeletal modelling, as well as surface electromyography, to conduct a comprehensive analysis of the working postures, muscular efforts and biomechanical loads that supermarket workers are subjected to during their daily work. A methodology for musculoskeletal modelling of MMH based on field measurements was developed and evaluated (Paper I), and hereafter used to perform a two-part risk assessment in two supermarkets (Paper II and III). In addition, musculoskeletal models were used to determine the effects of well-known lifting factors on the dynamic loading of the knees, shoulders and lumbar spine in a laboratory setting (Paper IV). Based on the field-based analysis, several MMH tasks that may pose a risk for the development of work-related musculoskeletal disorders were identified, while a large proportion of the analyzed tasks involved undesirable working postures. Based on the laboratory study, the lifting factors contributing most substantially to the dynamic loading of the involved joints were identified and the results compared to previous modelling studies of MMH as well as the field-based estimates. The dissertation was the first to employ state-of-the-art musculoskeletal models for the analysis of work-related MMH on a large scale, highlighting the potential of these models for improving our understanding of the dynamic loading of the involved joints.

DANSK RESUME

Muskel- og skeletbesvær er de mest udbredte arbejdsrelaterede helbredsproblemer og indebærer omfattende omkostninger for samfundet. Lænderygsmerter er den mest udbredte arbejdsrelaterede lidelse og udgør det største helbredsproblem på verdensplan målt i antal år med funktionsnedsættelse. Manuelt løftearbejde er den mest veldokumenterede risikofaktor for at udvikle lænderygsmerter. Manuelt løftearbejde er en fast del af arbejdet i adskillige brancher, heriblandt supermarkedssektoren. Få studier har dog undersøgt den fysiske belastning under arbejdet i supermarkeder på trods af at muskel- og skeletbesvær er udbredt i branchen. Denne ph.d.-afhandling anvendte avancerede metoder til inertibaseret bevægelsesanalyse, muskelskeletal modellering og elektromyografi i en omfattende analyse af arbejdsstillinger, muskel- og ledbelastninger under vareopfyldning i supermarkeder. En metodologi til muskelskeletal modellering af manuelt løftearbejde baseret på feltmålinger blev udviklet og evalueret (Paper I), hvorefter den blev anvendt til at foretage en todelt risikovurdering i to supermarkeder (Paper II and III). Herudover, så anvendtes lignende muskelskeletale modeller til at bestemme effekten af velkendte løfteforholdsfaktorer på de dynamiske ledkræfter i knæ, skuldre og lænderyggen baseret på laboratoriemålinger. Ud fra feltmålingerne blev adskillige arbejdsopgaver identificeret, som kan udgøre en risiko for udviklingen af muskel- og skeletbesvær, mens en stor del af de analyserede arbejdsopgaver involverede uhensigtsmæssige arbejdsstillinger. På baggrund af laboratoriestudiet blev de løfteforholdsfaktorer, som bidrog mest til den dynamiske belastning af de involverede led identificeret. Disse resultater blev sammenlignet med tidligere modelleringstudier af manuelt løftearbejde samt resultaterne fra feltstudiet. Dette var den første ph.d.-afhandling, som anvendte avancerede muskelskeletale modeller til en så omfattende analyse af manuelt løftearbejde og fremhævede herigennem potentialet af denne type modeller til at fremme vores forståelse af den dynamiske belastning af de involverede led.

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LIST OF STUDIES

Paper I

Larsen, F. G., Svenningsen, F. P., Andersen, M. S., de Zee, M. & Skals, S. Estimation of spinal loading during manual materials handling using inertial motion capture. *Annals of Biomedical Engineering* 2020;48;805-821.

Paper II

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Paper IV

Skals, S., Bláfóss, R., de Zee, M., Andersen, L. L. & Andersen, M. S. Effects of load mass and position on the dynamic loading of the knees, shoulders and lumbar spine: a musculoskeletal modelling approach. *Applied Ergonomics* (submitted).

ABBREVIATIONS

MSD = Musculoskeletal disorders

WRMD = Work-related musculoskeletal disorders

MMH = Manual material handling

NIOSH = National Institute of Occupational Safety and Health

A-P = Anteroposterior

sEMG = Surface electromyography

M-L = Mediolateral

AMS = AnyBody Modeling System

JRFs = Joint reaction forces

IMC = Inertial-based motion capture

GRF&Ms = Ground reaction forces and moments

IMUs = Inertial measurement units

OMC = Optical motion capture

MVIC = Maximal voluntary isometric contraction

LM = Load mass

AA = Asymmetry angle

HL = Horizontal location

VL = Vertical location

OMC-MGRF = Optical motion capture with measured ground reaction forces

OMC-PGRF = Optical motion capture with predicted ground reaction forces

IMC-PGRF = Inertial motion capture with predicted ground reaction forces

AMMR = AnyBody Managed Model Repository

%BW = Percentage of body weight

%BW x BH = Percentage of body weight times body height

A-C = Axial compression

nEMG = Normalized electromyography

ICC = Intraclass correlation coefficient

RMSE = Root-mean-square error

rRMSE = Relative root-mean-square error

R^2 = Coefficient of determination

PREFACE

The dissertation was a collaborative project between the National Research Centre for the Working Environment in Copenhagen, Denmark and Aalborg University, Aalborg, Denmark. The funding for the project was provided by the Independent Research Fund Denmark under grant no. DFF-7026-00099 to Sebastian Skals. Sebastian was enrolled in the Doctoral School in Biomedical Science and Engineering under the Faculty of Medicine at Aalborg University, but was based at the National Research Centre for the Working Environment in Copenhagen.

During the project period, Sebastian completed a 4-month research stay abroad at the Laboratory of Biological Structures Mechanics, IRCCS Istituto Ortopedico Galeazzi in Milan, Italy under the supervision of Dr. Fabio Galbusera and Dr. Tito Bassani. The work performed during this stay involved the implementation and evaluation of a novel musculoskeletal model of the thoracic spine. Unfortunately, due to a number of challenges with the model implementation as well as other parts of the project requiring more time than expected, this study was unfortunately not completed and included in the dissertation.

The novelty of the dissertation can partly be attributed to the composition of institutes and researchers, which brought together many areas of expertise, e.g. musculoskeletal disorders and physical workload, biomechanics and musculoskeletal modelling. Musculoskeletal models have great potential for the assessment of physical workload during manual handling, but has traditionally been limited by the overall complexity of the modelling process as well as the difficulty of acquiring the experimental input data. However, recent advancements have enabled the acquisition of these data outside a laboratory environment, making it possible to analyze e.g., manual material handling in more real-life scenarios. It was largely inspired by these developments that this project was formulated, as this technology could now potentially be used as a risk assessment tool in industrial settings. To test its applicability, the supermarket sector was chosen as the industry case study, as it has a high prevalence of manual material handling and musculoskeletal disorders, but has received little attention in the scientific literature. It was always prioritized that the results of the dissertation not only provided novel scientific contributions, but also produced valuable and useful results that could help reduce the incidence of work-related musculoskeletal disorders for supermarket workers.

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CHAPTER 1. INTRODUCTION

Musculoskeletal disorders (MSDs) are some of the most widespread and costly work-related health problems. Due to the complexity and multifactorial nature of work-related MSDs (WRMD), it has been challenging to determine the causative occupational exposures. Manual material handling (MMH) may be one of the most well-established risk factors for developing WRMD, particularly to the lower back. MMH is an integral part of the daily work in supermarkets and may be a contributing factor to the high prevalence of WRMD. Assessing musculoskeletal load during MMH and how these loads may lead to pain or injury have been major areas of biomechanical and ergonomics research. These efforts have involved a wide variety of methods with direct measurements and load estimates based on biomechanical models playing an important role. Today, scientific and technological developments have enabled the use of advanced musculoskeletal models to estimate internal forces during MMH in both laboratory and field settings. These types of models are able to estimate postures and motions as well as the forces in muscles and joints simultaneously, but have yet to be used for a comprehensive analysis of MMH.

In the following, the theoretical background of the doctoral dissertation is presented. First, the prevalence and societal costs of WRMD as well as the risk factors for developing these disorders are described. Second, the potential link between MMH and low back disorders is described, including a brief summary of the biomechanics of the lumbar spine. Third, the primary biomechanical methods used for assessing the load on the lumbar spine during MMH are presented in detail, while studies on load assessment in the shoulders and knees are briefly summarized. Fourth, a general description of musculoskeletal models is presented with particular emphasis on some recent advancements, which has provided new opportunities in the context of MMH analysis. Fifth, the prevalence of MSDs and MMH in supermarkets are described, as this sector was the primary focus of the dissertation. Finally, the overall aims of the dissertation are formulated with an overview of the included journal papers.

1.1. WORK-RELATED MUSCULOSKELETAL DISORDERS

The World Health Organization defines WRMD as health problems related to the muscles, tendons, skeleton, cartilage, ligaments and nerves, which are induced or aggravated by work and the circumstances of its performance [1]. For a MSD to be work-related, either the work environment or performance of the work have to contribute significantly to the development or persistence of the condition [2]. However, WRMD are multifactorial in nature and associated with physical, psychosocial and individual factors [3,4]. Hence, it is challenging to determine the causal pathway from occupational exposure to a MSD, as many factors may contribute to the condition simultaneously, which may or may not be work-related. For this reason, the science of occupational health is likewise multifactorial and dependent on a diverse set of scientific disciplines to be able to provide meaningful recommendations for the regulation of work [3]. For example, epidemiology is used to determine the prevalence of disease in a working population and how the occurrence of a particular disease may coincide with the occurrence of occupational or non-occupational risk factors. Biomechanical, physiological and psychophysical methods are used to explore the relationship between physical workload and the tolerance of the loaded biological structures, while psychological methods may be used to identify contributory mental and organizational factors. Finally, the resulting evidence-based recommendations need to be implemented nationally or locally, which require organizational, financial and political considerations.

1.1.1. PREVALENCE AND SOCIETAL COSTS

Due to the diversity and complexity of MSDs as well as the use of many different classifications to describe these conditions, it is challenging to provide accurate overall estimates of the prevalence and cost of MSDs in society. For example, Roquelaure [5] presents an overview of the 26 main periarticular diseases that are generally classified as MSDs (e.g. tendinopathies, tunnel syndromes and nerve compressions) as well as the main non-specific MSDs in the limbs and spine, such as non-specific pain in the upper limbs, cervical and lumbar spine. As no classification scheme has been universally adopted, it limits the ability of clinicians and researchers to communicate in a consistent and meaningful way [4]. The task is further complicated by the fact that workers tend to significantly underreport MSDs: for example, Riviére et al. [6] found that MSDs in the shoulders, elbows and lumbar spine were not reported between 63% and 73% of the time across 10 regions in France. Then there is the multifaceted nature of the societal costs, such as sickness absenteeism and presenteeism [7], lost productivity as well as acute and long-term medical care [8]. Despite these challenges, efforts have been made to estimate the overall impact of MSDs in society.

In the *Sixth European Working Conditions Survey* from 2015, the most reported health problem for workers in the European Union was backache (43%), followed by muscular pain in the neck or upper limbs (42%), headache and eyestrain (35%), overall fatigue (35%) and muscular pains in the hip or lower limbs (29%) [9]. Chronic musculoskeletal pain was also widespread, although much lower, with 11% and 16% reporting chronic neck and low back disorders, respectively. Another interesting finding from this report was that of the workers who mentioned that they suffered from any work-related health problem, 60% listed MSDs as the most serious, while “stress, depression and anxiety” was the second most mentioned (16%) [10]. As described by Bevan [8], ad hoc analysis of the *European Labour Force Survey* has shown that MSDs accounted for 53% of all work-related diseases, 50% of all absences from work lasting more than three days, 49% of all absences lasting two weeks or more and roughly 60% of all reported cases of permanent incapacity across 15

European countries. Overall, the total yearly cost of WRMD was estimated to around €240 billion or up to 2% of gross domestic product. Globally, MSDs also constitute some of the most widespread and costly health problems. In 2015, low back and neck pain was listed as the number one health problem in the world measured in years lived with disability [11], mirroring the results from 2005 and 1995. In Denmark, the prevalence of people living with low back and neck pain in 2010 were approximately 16% and 10%, respectively, with an estimated total cost of roughly €368 million for treatment and €922 million in lost production [12].

Although these global and national estimates include all instances of back and neck pain in the general population, some efforts have been made to determine the proportion of cases that are caused by occupational factors. For example, Guo et al. [13] collected questionnaire data from approximately 30,000 workers in the United States and found that the workers reported over 5000 back pain cases over a 1-year period, corresponding to a prevalence of 17.6%. Of these cases, approximately 65% were related to combined occupational exposures. Based on a review of epidemiological studies, Punnet et al. [14] estimated that 37% of global low back pain cases were associated with occupational exposures, resulting in 818,000 years lost to disability (or disability-adjusted life years lost). Furthermore, the U.S. Bureau of Labor Statistics recorded 900,380 non-fatal occupational injuries involving days from work across the United States private industry in 2018; 282,860 of these cases were due to “overexertion and bodily reaction” [15].

In view of the above, it is clear that the societal costs of MSDs are tremendous and occupational factors contribute greatly to the development and persistence of these disorders.

1.1.2. RISK FACTORS

Epidemiological studies have identified several occupational factors that may cause or contribute to the development of WRMD. In accordance with the scope of the dissertation, only the occupational risk factors that have been associated with the development of knee, shoulder and low back disorders are presented here.

The most common risk factors associated with knee disorders are exposure to lifting [16-19] and kneeling [16,18-20]. The combination of kneeling and heavy lifting in particular, has shown strong associations with the development of knee disorders [18,19]. In a review of longitudinal studies, Da Costa and Viera [17] also found evidence for an association between knee disorders and repetitive work and awkward postures. In general, there is less evidence for the work-relatedness of lower-extremity disorders compared with upper-extremity and back disorders, possibly due to the comparatively lower prevalence [21]. A review by Reid et al. [18] suggest that physically demanding jobs, such as construction, forestry and farm work are the professions typically affected by serious knee disorders, such as osteoarthritis.

Some of the most common work factors contributing to shoulder disorders are working with the arms elevated [2,22-26], frequent handling of loads [24], repetitive work [2,24], heavy physical work [2,17], manual material handling [23], bending and twisting [23] as well as the combination of overhead work with other exposures [26]. Based on these studies, it seems that the identification of risk factors for shoulder disorders are limited by the use of different terminology for both the occupational risk factors and type of shoulder disorders (e.g. subacromial impingement syndrome [24] and debilitating/non-debilitating pain and supraspinatus tendinitis [25]). Despite this limitation, the literature seems to support a positive association between shoulder disorders and heavy physical work, manual material handling and repetitive work, while the associations to working with arms elevated has been particularly well established. For example, Punnet et al. [22] found that severe shoulder flexion or abduction, especially for more than 10% of the work duration, to be predictive of chronic or recurrent shoulder disorders in autoassembly workers. Shoulder disorders were the most common musculoskeletal problem reported for these workers, resulting in substantial costs due to medical treatment, lost time from work and work restrictions. Svendsen et al. [25] found that a 1% increment in the duration of daily working hours involving arm elevation above 90° showed odds ratios of 1.23 for supraspinatus tendinitis, 1.16 for debilitating shoulder pain and 1.08 for non-debilitating shoulder pain in a cohort of male machinist, car mechanics and house painters.

Low back disorders have been associated with several occupational risk factors, such as manual material handling or lifting [2,17,21,27-34], awkward postures [2,17], bending and twisting [21,27,28], and heavy physical work [2,17,21,33]. However, due to methodological limitations, most epidemiological research has failed to establish causality between any of the above-mentioned risk factors and low back disorders [21]. This limitation is further highlighted in a recent review of systematic reviews by Swain et al. [33], which showed positive associations for many occupational risk factors, but no consensus on causality across reviews. However, the authors found that the strongest support for a positive association with low back pain was for heavy physical work, including manual material handling and lifting.

Studies involving more detailed exposure assessments tend to more strongly support the association between MMH and the incidence of low back disorders [21,35]. For example, Punnet et al. [36] and Marras et al. [37] used case-control designs with detailed postural assessments to determine the quantitative relationship between trunk motion and work-related low back disorders. Punnet et al. [36] found odds ratios of 4.9, 5.7 and 5.9 for mild trunk flexion, severe trunk flexion and trunk twist or lateral bending, respectively. Marras et al. [37] showed an odds ratio of 10.7 for high risk compared with low risk jobs, classified using a combination of motion and workplace factors, such as lifting frequency and trunk kinematics. These studies are some of the earliest examples of using a combination of biomechanical variables and injury records to predict the amount of exposure to biomechanical risk factors that may result in work-related low back disorders.

1.1.3. MUSCULOSKELETAL INJURY CAUSATION

The mechanisms that cause overexertion injuries to muscles, tendons and ligaments are still not completely understood [38]. Based on a comprehensive review of the literature, Kumar [39] proposed four theories of musculoskeletal injury causation: 1) the Multivariate Interaction Theory, 2) Differential Fatigue Theory, 3) Cumulative Load Theory and 4) Overexertion Theory. The multivariate interaction theory simply highlights the multifactorial nature of musculoskeletal injuries, meaning that one needs to consider injury causation as an interactive process between biological, psychosocial and biomechanical factors that each contain many potentiating variables. In other words, the individual's biological characteristics and psychosocial profile affect their response to biomechanical stresses, hereby modulating the final injury and pain outcome. The differential fatigue theory refers to the consequences of occupational demands being prioritized over the compatibility between task demands and the workers' physical capacity. Hence, occupational activities are often highly repetitive to increase their economic value and involve a large number of muscles at various joints. The differential and repeated loading of the joints and muscles may lead to different amounts of muscle fatigue, which in turn, lead to altered joint kinematics and non-optimal loading patterns. The cumulative load theory refers to the fact that biological tissues undergo mechanical degradation with prolonged usage due to their visco-elastic properties, while cumulative fatigue may also reduce their stress-bearing capacity. These changes can reduce the threshold stress at which the tissues fail. Finally, the overexertion theory implies that the physical effort required for a task may exceed the tolerance limits of the biological structures and is a function of force, posture, motion and duration. A musculoskeletal injury will likely be a result of an interaction of variables across all these four theories [39].

1.2. MANUAL MATERIAL HANDLING AND LOW BACK DISORDERS

Based on the epidemiological research, it is evident that the occupational risk factor with the strongest and most consistent association to the development of low back disorders is MMH. The term MMH typically refers to the acts of lifting, lowering, pushing, pulling, holding and carrying materials [40]. During handling operations, the body experiences both internal stresses, resulting from internal pressure, tension in the surrounding muscles and passive structures, and external stresses from the weight of the body segments and the handled load [32]. In some cases, an acute overexertion can cause a muscle rupture or fissure in the intervertebral discs, but more often a triggering event may result in micro trauma to the tissues leading to degeneration [38,41]. However, the accumulation of trauma will be more rapid with higher loads [41]. As described by Bazrgari and Xai [42], two main pathways between lower back loading and the development of low back pain have been established: 1) tissue failure or nerve irritation due to acute excessive mechanical loading, and 2) the accumulation of micro trauma due to cumulative loading, which may decrease the threshold of tissue failure or nerve excitation due to e.g. muscle fatigue or creep deformation of the passive tissues. Even though other factors, e.g. psychosocial, are believed to affect several aspects of work-related low back pain, an injury must first result from excessive mechanical loading, as it is the characteristics of the load itself and the properties of the tissue that determine the type and extent of damage [41]. However, when injury or pain is present, psychosocial factors may modulate and exacerbate the experience and persistence of the trauma [43].

Task-based analysis of MMH has traditionally involved a diverse set of methods based on biomechanical, physiological and psychophysical approaches [40]. These methods range in complexity from self-reports and observational methods to direct measurements and biomechanical models [44-47]. The choice of method typically involves a trade-off between complexity and cost, as ergonomist are often expected to solve injury problems in a way that require minimal capital investment [48]. Some of the earliest examples of the biomechanical approach has been the work by Chaffin et al. [49-52], which incorporated a sagittal-plane biomechanical model to estimate the strength requirements of industrial handling jobs and compared these estimates with human strength capabilities. The model of Chaffin et al [49,53] would later be used by the National Institute of Occupational Safety and Health (NIOSH) in their *Work Practices Guide for Manual Lifting* [54]. By additionally incorporating failure tolerance data of cadaver discs [55,56], tolerance limits of 6400 (maximum permissible limit) and 3400 N (action limit) were proposed for the compressive force at the L5-S1 joint. Perhaps the most well-known application of the psychophysical approach is the work by the Liberty Mutual Research Center (e.g. Snook and Irvine [57], Snook [58] and Snook and Ciriello [59]). These studies used a combination of physiological measurements (e.g. oxygen consumption and heart rate) and subjective assessment of perceived exertion as a basis for recommending maximum acceptable weights in lifting, lowering, pushing and pulling [59]. Examples of the physiological approach include studies by Garg et al. [60] and Garg and Saxena [61], which used metabolic and heart rate to determine the physiological efforts of manual lifting in a laboratory and field setting, respectively. Among these three overall domains for studying MMH, the biomechanical approach is the only one incorporating an explicit hypothesis of an injury mechanism, such as the 3400 N spinal compression action limit recommended by NIOSH [48].

In the following, the biomechanical approach to determine the injury risk to the lumbar spine during MMH is summarized with particular emphasis on the use of biomechanical models.

1.3. BIOMECHANICS OF THE LUMBAR SPINE

The functional anatomy and biomechanics of the spine is a vast and intricate area of research. Hence, the purpose of this section is simply to provide an overview of the basic structures of the spine, particular the lumbar region, and hereby, provide the terms and concepts that are meaningful to the subsequent discussion of low back loading.

The whole spine consists of the cervical, thoracic, lumbar, sacral and coccygeal vertebrae as well as the intervertebral discs, ligaments, rib cage and spinal muscles (see Fig. 1.1) [62]. As described by Galbusera and Wilke [63], the cervical spine has seven vertebrae (C1-C7) and its primary function is to provide mobility to the head. The thoracic spine consists of 12 vertebrae (T1-T12) and is the main support of the ribcage. The lumbar spine consists of five vertebrae (L1-L5) and provides a substantial proportion of the trunks mobility as well as being subjected to the highest loads. As described by Bogduk et al. [64], the sacrum consist of several fused vertebrae lying at the base of the vertebral column, wedged between the two iliac bones, forming the posterior wall of the pelvis. Hence, all longitudinal forces affecting the lumbar spine are also transmitted to the sacrum, while its position in the pelvic girdle enables it to transmit forces from the vertebral column in the transverse directions to the lower limbs and vice versa.

In the lumbar region of the spine, there are three overall groups of muscles: 1) psoas major, 2) intertransversarii laterales and quadratus lumborum, and 3) the lumbar back muscles [64]. As described by Bogduk [64], the psoas major, intertransversarii laterales and quadratus lumburom provides minor contributions to spinal motion, but may exert high compressive forces on the lower lumbar discs, provide feedback from the movements of the spinal column influencing the action of the surrounding, larger muscles and contribute to the movement of the 12th rib during respiration, respectively. The lumbar back muscles include the interspinalis, intertransversarii mediales, multifidus and erector spinae longissimus thoracis and iliocostalis lumborum. Overall, the lumbar back muscles provide many possible actions in response to the movements of the spinal column, including minor active movements, postural adjustive movements and major movements in forward bending and lifting. Whenever the back muscles contract, they exert longitudinal compression, which raises the pressure in the intervertebral discs. The muscles and tendons are the means through which the spinal system generates forces and are instrumental in providing spinal stability [65].

The ligaments of the lumbar spine consists of the ligaments connecting the vertebral bodies (i.e. the annuli fibrosis, anterior and posterior longitudinal ligaments), the ligaments of the posterior spinal elements, the iliolumbar ligament and so-called false ligaments [64]. Collectively, these passive tissues provide stability to the spine with each ligament resisting a specific motion direction [66]. The ligaments do not provide

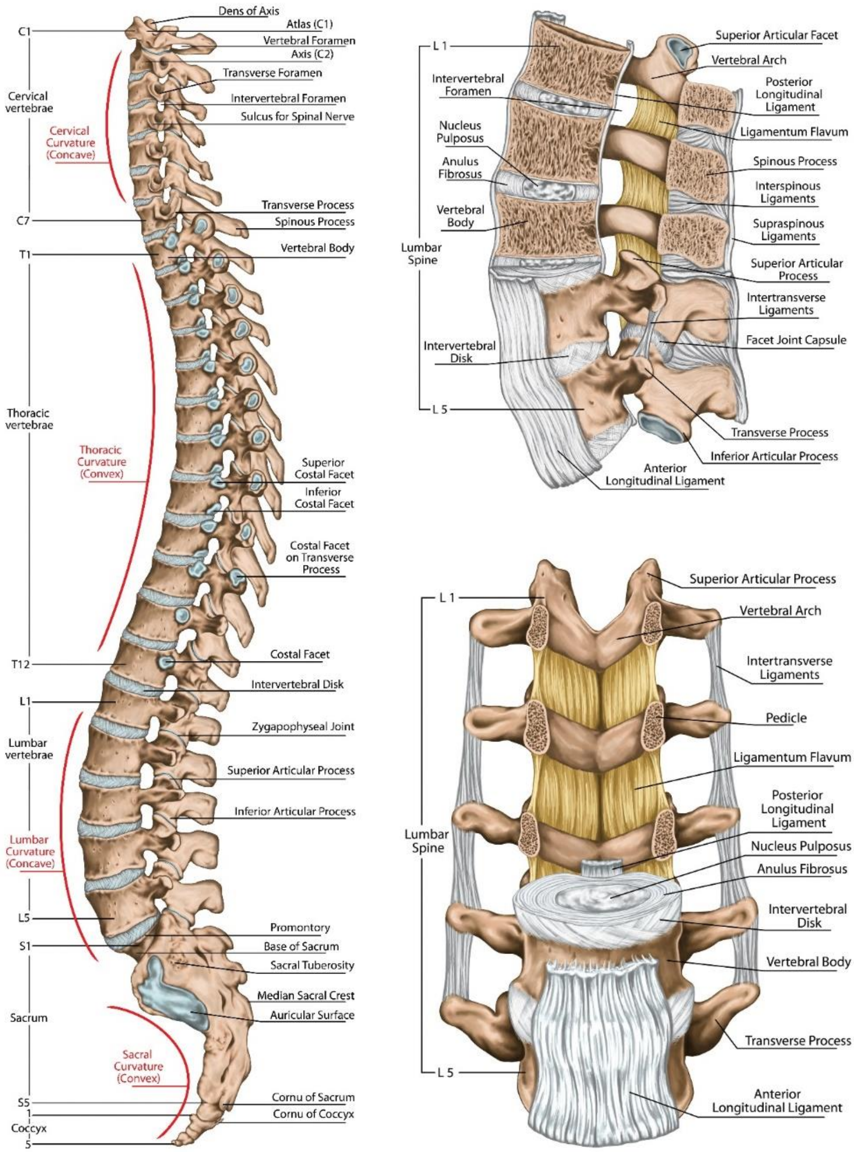

Figure 1.1 – Left: illustration of the whole spine, including the cervical, thoracic and lumbar spine as well as the sacrum and coccyx. Right: Detailed lateral (top) and anterior view (bottom) of the lumbar spine. Source: shutterstock.com/stihii

significant stability near neutral posture, but develop reactive forces that resist spinal motion toward the ends of the ranges-of-motion [65].

The basic building blocks of the spine are called functional spinal units, which consists of two adjacent vertebrae, the intervertebral disc, the zygapophysial joints (also called facet joints) and the spinal ligaments [62]. As described by Wilke and Volkheimer [66], the vertebral body is the main weight-bearing structure with a shell of compact bone, which are reinforced by the vertical and horizontal trabeculae. The posterior structures of the vertebrae are connected to the vertebral bodies via two short pillars, called the pedicles, which extend from the posterior wall and consists of spongious bone covered by a shell of compact bone. The pedicles provide attachment for the posterior spinal structures, which are formed by the laminae and the spinal processes; the transverse, accessory, mammillary and spinous process serve as attachment for the musculature, while the articular processes constitute synovial joints that contribute to the load sharing between the anterior and posterior spinal columns. These synovial joints are called the zygapophysial joints, which are formed by the articulation of the inferior articular processes of one lumbar vertebrae with the superior articular processes of the next vertebrae. These joints are important for preventing forward displacement and rotatory dislocation of the intervertebral joint [64]. As described by Wilke and Volkheimer [66], the intervertebral joint is a flexible fibrocartilaginous joint connecting the vertebral bodies, also called the intervertebral discs, which consist of the central nucleus pulposus surrounded by the concentric annulus fibrosus. In addition, two layers of cartilage cover the top and bottom of the disc, called the vertebral endplates, which separates the discs from the adjacent vertebral bodies. The main functions of the disc are to allow movement between vertebral bodies and to transmit loads from one vertebral body to the next [64].

1.4. ASSESSMENT OF SPINAL LOADS

Assessment of low back loading has traditionally involved in vivo measurements, in vitro testing of cadaver specimens as well as load estimates based on biomechanical models. Finite element models have also been widely used to study the biomechanics of the lumbar spine (see e.g. Schmidt et al. [67] and Fagan et al. [68]), but are beyond the scope of this dissertation. In the following, the other three approaches to study low back loading are summarized.

1.4.1. IN VIVO

There are mainly two types of in vivo measurements that have been used to assess low back loading, namely intradiscal pressure measurements [69-72] and telemeterized vertebral body replacements [73]. Both of these methods are limited by the ethical complications of inserting measuring devices inside the body and are therefore, extremely rare. Pressure needles were first used in a series of experiments in the 1960s and 70s to estimate the intradiscal pressure during various activities, e.g.

by Nachemson et al. [74,75]. Later work by Wilke et al. [70,71] used this technique in a healthy subject during a series of postures and lifts, while Sato et al. [72] studied subjects with various spinal pathologies. The work of Wilke et al. [70,71] was published along with the anthropometric data of their subject and has since been used to evaluate the accuracy of computational models [47,76]. Spinal loads based on vertebral body replacements during various activities of daily living [77] and lifting [78] has also been published, but these studies are limited by the fact that much of the load in the spine is transferred to the implanted spinal fixation device and additional structures [73].

1.4.2. IN VITRO TESTING AND INJURY TOLERANCE LIMITS

In vitro studies have been essential to our understanding of the mechanical tolerance of single vertebra, intervertebral discs and functional spinal units [79]. In short, this method involves using mechanical testing devices to subject cadaver specimens to various loads, such as pure compression [80], sagittal bending [81] or complex loading patterns, e.g. simultaneous compression, flexion, lateral bending and axial torsion [82]. Using this approach, the loading patterns can be systematically tested to determine at which load, rate and duration the specimens are damaged. As described by Cruz et al. [83], there are both advantages and disadvantages to in vitro testing. One of the most important advantages is that this method provides the ability to determine at what point in a loading cycle injury occur. However, testing cadaver specimens is expensive, as they can only be used once, and they lack true in vivo characteristics, meaning that they are not necessarily reflective of the load tolerance in living beings. For example, cadaveric tissues are affected by temperature changes, which may change the extensibility of tendons and ligaments, while the capacity for self-healing of biological tissues is lost in this non-physiological state [84]. Furthermore, there is large variability in the load tolerance between specimens, as their strength is dependent on e.g., age, gender, body mass and the spinal level tested [85].

Despite these limitations, in vitro studies have been essential for determining the critical loads that may result in low back injury. For example, two comprehensive reviews of in vitro studies by Genaidy et al. [85] and Gallagher and Marras [84] provide well-founded tolerance limits for compression and shear loading of the lumbar spine, respectively. Genaidy et al. [85] emphasized the idea of setting the limit for compressive strength based on the concept of damage load to the functional spinal unit, i.e. the weight that causes the first gross signs of damage, while adjusting the compressive strength for age, gender and body weight. Based on the proposed equation and the damage load estimates of 33 to 93% of compressive strength found in Eie [86], the damage load for men and women in the age group 20-29 years can be calculated to 3268 and 2314 N. However, using the less conservative estimate of damage load (82% of compressive strength on average) found in a more recent in vitro study by Yoganandan [87], the corresponding injury thresholds would be 4480 and

3431 N. In this case, the latter threshold value for women of 3431 N is the same as the action limit proposed by NIOSH [54,88]. Gallagher and Marras [84] reviewed in vitro fatigue failure data due to anteroposterior (A-P) shear loading and proposed a criteria of 1000 N for infrequent loading (< 100 loadings per day) and 700 N for frequent loading (100-1000 loadings per day), which would be protective for 90% of individuals. These criteria generally support the earlier recommendations by McGill [89], who proposed a maximum permissible limit of 1000 N and an action limit of 500 N. As can be seen from these recommendations, the tolerance limits for shear loading are much lower than for compression. However, as described by Gallagher and Marras [84], the spinal structures that are loaded in shear are also weaker. In particular, the collagen fibers in the intervertebral discs are poorly oriented to resist shear, so much of the resistance stems from the neural arch, zygapophyseal joints and spinal ligaments. During MMH, shear forces typically occur as a result of high degrees of torso flexion, which limits the ability of the back extensors to resists anterior shear [41].

It is important to note that the tolerance limits for critical low back loading are controversial. The most widely used criteria are the NIOSH recommendations. Two in vitro studies [55,56] as well as studies linking predicted static compressive forces on the L5-S1 disc with lifting-related low back pain (e.g. Chaffin and Park [51] and Herrin et al. [90]) were instrumental for formulating these recommendations, but the criteria are still largely based on expert consensus [88]. In a critical review, Jäger and Luttman [91] states that the NIOSH action limit is neither epidemiologically nor biomechanically supported by these foundational studies. However, 40 years after it was first proposed, the NIOSH criterion of 3400 N still appear to be the best available estimate of critical low back loading.

1.4.3. BIOMECHANICAL MODELS

Due to the ethical and methodological complications of applying in vivo and in vitro measurements, biomechanical models have played a major role in determining the loads on the lower back during MMH. Many different types of models with varying complexity and versatility have been developed over the years, as for instance, 2-D static and 3-D dynamic models as well as models based on surface electromyography (sEMG). Morris et al. [92] made one of the first attempts to calculate the load on the lower back during lifting using a 2-D static biomechanical model. This model was further developed by Chaffin et al. [49,53] and used in a series of studies to estimate the load on the back during sagittal-pane lifting [50-52]. As mentioned previously, this work was foundational for developing the original NIOSH guidelines for manual lifting [54]. The guidelines were later revised by Waters et al. [88], which led to the development of *The Revised NIOSH Lifting Equation*. This equation has since been shown to be reasonable predictor of the risk of low back injury during lifting [93,94].

However, there are a number of important limitations associated with the use of 2-D static biomechanical models. First, several studies have shown that these models underestimate the moments and compressive forces in the lower back [95-98]. This is due to the fact that the acceleration components and inertial forces of the load and body segments are ignored in static analysis [99]. Second, most of these earlier examples of static models do not include representations of the lumbar back muscles, meaning that the influence of muscle co-contraction on the spinal compression and shear forces were not considered [38,100]. For example, Granata and Marras [100] found that neglecting muscle co-activity resulted in an underestimation of the compression and shear forces in the lumbar spine during dynamic lifting by 45 and 70%, respectively. Finally, 2-D models do not account for the potential influence of load asymmetry, which may also affect the compression, A-P shear and mediolateral (M-L) shear forces in the lumbar spine [101-103].

To overcome these limitations, 3-D dynamic sEMG-assisted biomechanical models have been developed [104-106], which accounts for muscle co-activity and 3-D lifting dynamics. However, sEMG-assisted models require detailed kinematic data as well as sEMG-measurements of multiple muscles to be able to estimate the joint loads, which have traditionally prohibited their use outside a laboratory environment. This limitation also applies to 2-D and 3-D dynamic models that do not incorporate sEMG-measurements, as for instance, the models by de Looze et al. [97] and Kingma et al. [107], which still require measurements of kinematics and external forces, e.g. ground reaction forces. Therefore, the use of both dynamic and sEMG-assisted biomechanical models in industrial settings have traditionally been infeasible due to the difficulty of acquiring the necessary input data [38].

1.5. ASSESSMENT OF LOADS IN THE KNEES AND SHOULDERS

In the context of MMH, much less attention has been given to the loads in the shoulders and knees compared with the lower back. This is likely because musculoskeletal disorders in the knees and shoulders are less commonly associated with handling activities.

Probably the most common method used to study the load on the shoulders during MMH have been sEMG [108-113], which is typically used to study the relative activation of the surface musculature during different tasks. 2-D static [114] and dynamic [115] as well as 3-D dynamic biomechanical models [116] have also been used to estimate shoulder moments during MMH. More complex shoulder models have also been developed, which provide a much more detailed representation of the shoulders functional anatomy [117-119]. These musculoskeletal models are based on inverse dynamic optimization to distribute the muscle and joint forces. For example, Hoozemans et al. [120] used the model of van der Helm [117] to study shoulder loads during pushing and pulling, but these types of models have otherwise had limited use for the analysis of MMH.

Studies of the loads in the knee joints during lifting are also rare. de Looze et al. [121], Schipplein et al. [122,123], Delisle et al. [124] and Lavender et al. [125] represent some of the few examples in the literature, which have evaluated the loads on the lower extremities in addition to the lower back during MMH. Although these models were relatively simplistic (2-D and 3-D dynamic models without muscles), they provide valuable information about the load sharing between the involved joints. As the complexity of the model architecture increases, the load sharing between joints can potentially be more accurately estimated. This is exemplified in a recent study by van der Have et al. [126], which used a full-body musculoskeletal model to calculate flexion-extension joint moments for the shoulder, L5-S1, knee and hip joints during stoop and squat lifting.

1.6. MUSCULOSKELETAL MODELS

As indicated above, muscle and joint forces have also been estimated using optimization-based musculoskeletal models. These models are based on the assumption that a cost function can be minimized, while maintaining dynamic equilibrium – most often the sum of muscle activities to different powers [73]. For example, constraint equations on the muscles may ensure that the forces they produce are positive and within the limits of their maximum strengths, while the muscle and joint forces are distributed by minimizing the sum of muscle activities [127,128]. By using optimization, the fundamental problem of muscle redundancy can be solved, meaning that there are more muscles available than necessary to drive the body's degrees-of-freedom [127]. This approach is founded on the idea that the central nervous system attempts to find the most optimal solution for producing a given motion, e.g. that the overall load on the muscles and body are minimized [128].

Over the past decades, scientific and technological advancements as well as the availability of commercial modelling and simulation software, e.g. OpenSim [129] and the AnyBody Modeling System (AMS) [127], has made this type of analysis more readily available to science and industry [130]. These models now provide valuable information to a diverse set of scientific fields, as for instance, clinical gait analysis [131], orthopedics [132,133] and ergonomics [134,135].

It is generally recognized that the more accurately the models represent the musculoskeletal system (i.e. joint definitions, muscle geometry, passive tissues etc.), the more likely it will be that the estimates of internal forces will be valid. In the AMS, the body parts have been developed independently by various research groups over the last decades and are largely based on cadaver datasets, as for instance, the lower extremity [136] and shoulder and arm model [137-139]. The models of the various body parts have been integrated into a full-body model with upwards of a 1000 muscle elements [140], providing a powerful tool to accurately estimate muscle and joint reaction forces (JRFs). Being able to estimate the forces in multiple joints simultaneously is one of the great advantages of the AMS. Generally, comparative studies also show that the AMS model provide very reasonable estimates of in vivo forces. For example, studies by Bassani et al. [76] and Rajae et al. [47] found that the AMS model's estimates of the compressive forces in the lumbar spine were in close agreement with the intradiscal pressure measurements of Wilke et al. [2001]. Furthermore, Rajae et al. [47] showed that the AMS model was superior to other common lifting analysis tools in this regard. Despite these promising results, it remains a great challenge to validate the results of musculoskeletal models, as in vivo data are rare and mostly available in the form of instrumented joint replacements (e.g. Bergman et al. [141] and Rohlman et al. [77]). Therefore, trend validation has been recommended, which can help evaluate whether the model components interact correctly with each other by systematically changing parameters and monitoring the outputs as a function of these changes [130].

In view of the above, there is great potential in the application of advanced musculoskeletal models for assessing the load in the joints during MMH. However, similarly to other 3-D dynamic and sEMG-assisted models, acquiring sufficiently detailed experimental input data outside a laboratory environment has traditionally been infeasible. Most modelling studies have relied on marker-based motion analysis and force plate measurements to acquire these input data, which is expensive, time-consuming and highly restrictive in regards to the execution of the measured tasks.

1.6.1. RECENT ADVANCEMENTS

In recent years, two major advancements have provided new opportunities for applying the AMS for MMH analysis in a field setting, namely ground reaction force prediction and inertial-based motion capture (IMC) technology. First, methods for predicting ground reaction forces and moments (GRF&Ms) based on segment kinematics and dynamical properties only have been developed, which utilize dynamic contact elements under the feet [142-144]. These methods have shown comparable accuracy to force plate measurements during activities of daily living [142], sports-related movements [143] and inertial-based gait analysis [144]. Second, IMC technology, such as the Xsens MVN Link and Awinda systems [145,146], have enabled the acquisition of kinematic data outside a laboratory environment with sufficient detail to drive full-body musculoskeletal models [147]. The Xsens MVN Awinda system in particular, provides a setup of 17 inertial-measurement units (IMUs) attached with velcro straps, which is very suitable for application in the field, where ease-of-use and non-obstructiveness are essential. These systems have shown reasonable accuracy compared with marker-based motion analysis for tracking activities of daily living [148,149]. The combination of these methods may provide a powerful new tool to identify postural risk factors as well as estimate muscle and joint forces from detailed musculoskeletal models based on field-measurements.

1.7. SUPERMARKET SECTOR

Many major industries require a large amount of MMH. A study by Heran-Le Roy et al. [150] found that 51.6% of workers in retail trade – an umbrella category including the grocery or supermarket sector – were exposed to MMH with 24% of the exposed workers performing MMH more than 20 h per week. This was the second highest exposure among the 34 occupational categories surveyed. As there is a strong association between MMH and low back disorders, it is not surprising that the supermarket sector also has a high prevalence of back pain compared with other major industries [151]. In addition to back pain, supermarket workers generally report a high prevalence of several WRMD [152-154]. For example, Forcier et al. [154] found that musculoskeletal injuries accounted for 63% of all compensable injuries and 73% of days away from work with the most affected body regions being the lower back (37%), shoulders (16%) and wrists (9%). Anton and Weeks [153] found that 78% of supermarket workers reported some musculoskeletal symptom over a 12-month period, particularly to the lower back (51%), feet (50%) and shoulders (31%). In addition, data from the U.S. Bureau of Labor Statistics showed that there were 28,340 non-fatal occupational injuries and illnesses involving days away from work reported in the supermarket sector in 2018 [15]. Of these cases, 10,970 were due to “sprains, strains and tears”, and 3,300 due to “soreness and pain” with the back (4,870), shoulders (2,730) and knees (2,430) being some of the most frequently affected body regions. Furthermore, 10,230 of the total cases were related to “overexertion and bodily reaction” of which 4,860 were specifically linked to lowering and lifting.

Despite the high prevalence of WRMD in the sector, few studies have attempted to determine the causative occupational exposures. Most research employing measurements of physical workload or evaluating the efficacy of interventions has targeted the cashiers and checkstands [155-158]. However, most supermarket employees are primarily engaged in the receiving, stocking and re-arranging of groceries. MMH also appear to be the most frequently identified occupational risk factor for developing WRMD in the sector [159-163]. However, there is little data on the physical efforts required by these workers [111,162]. A few examples exists in the literature that have employed sEMG-measurements alone [164] or in combination with motion analysis [111,165] to assess the physical workload during stocking work. For example, Ohu et al. [164] showed a reduction of muscle activities in the lower back, arms and shoulders when a mobile cart was used for stocking, while Balogh et al. [165] showed that stocking work resulted in the highest trapezius muscle activity compared with cashier, mixed and delicatessen work. Both these studies were performed in the field, but do not provide sufficient information to identify potentially hazardous MMH tasks.

In view of the above, there is an urgent need for a comprehensive analysis of the physical efforts required to perform MMH in the supermarket sector in order to identify the tasks that may increase the risk of developing MSDs.

1.8. AIMS

The overall aims of the doctoral dissertation were to develop and evaluate a methodology for field-based analysis of MMH based on state-of-the-art musculoskeletal models, and apply these methods for a comprehensive analysis of MMH in the supermarket sector to assess the risk of developing handling-related MSDs. To determine the relative importance of well-known lifting factors on dynamic joint loads and provide context for the field-based analysis, the same models were also implemented for a laboratory-based study of MMH. Three experimental studies were carried out for this purpose, which formed the basis for four scientific papers (Paper I-IV). In Paper I, the methodology for field-based analysis of MMH using musculoskeletal models was evaluated by comparing the model estimates to those obtained from a more traditional laboratory-based approach. In Paper II and III, this methodology was implemented in combination with sEMG-measurements for a comprehensive risk assessment in two supermarkets, in which common MMH tasks were ranked according to postures, muscle activities and joint loads. In Paper IV, musculoskeletal models were used to determine the effects of load mass and position on multiple joint loads, hereby providing an assessment of the relative importance of well-known lifting factors as well as detailed reference data for field-based studies applying musculoskeletal models. Based on these studies, new insights may be gained on the physical efforts required to perform MMH in supermarkets and which tasks that may pose a risk for developing MSDs. Furthermore, the influence of lifting factors on multiple joint loads may provide an improved basis for advising about the regulation of MMH in general, while providing reference data to evaluate field-based estimates of musculoskeletal load. Finally, by discussing the strengths and limitations of the proposed methodology, the dissertation highlights its potential and the improvements necessary to apply this technology for the analysis of MMH on a larger scale.

CHAPTER 2. METHODS

In the following, the materials and methods used in the experiments forming the data foundation of the dissertation are summarized, while a detailed description can be found in the appended papers. The experiments associated with Paper I, II-III and IV are from heron referred to as Experiment I, II and III, respectively (see Table 2.1).

Table 2.1 – Overview of experiments and associated papers

Experiment	Paper	Purpose
I	I	Evaluation of methodology for field-based analysis
II	II and III	Risk assessment of manual material handling tasks in two supermarkets
III	IV	Laboratory study on the effects of ergonomic lifting factors on joint loads

2.1. SUBJECTS

All the conducted experiments used a cross-sectional design. For Experiment I and III, the subjects represented convenience samples, which primarily included a mixture of university students, colleagues, acquaintances and six supermarket workers. For Experiment II, supermarket workers were recruited with the assistance of the senior human resources specialist of the participating supermarket company. In short, this process involved contacting store managers throughout the North Jutland Region of Denmark and informing them of the aims and procedures of the experiments. Two store managers agreed to involve their stores from which 15 workers volunteered to participate in addition to the store managers themselves. In total, 13, 17 and 22 subjects participated in Experiment I, II and III, respectively (see Table 2.2). The studies followed the guidelines of the North Denmark Region Committee on Health Research Ethics and all subjects provided written informed consent. Data were collected between March 2018 and May 2019.

Table 2.2 – Subject information, including sample size (n), sex, age, mass and height

Experiment	n	Male/female	Age (years)	Mass (kg)	Height (cm)
I	13	9/4	26 ± 3	76.4 ± 12.8	179.3 ± 7.8
II	17	8/9	27 ± 8	76.6 ± 14.7	174.4 ± 9.1
III	22	16/6	30 ± 10	80.6 ± 12.1	178.1 ± 11.3

2.2. INSTRUMENTATION

Experiment I and III were conducted in a laboratory setting, specifically the Human Performance Laboratory at the Department of Health Science and Technology, Aalborg University. In Experiment I, the Xsens MVN Awinda wireless motion-tracker (Xsens Technologies BV, Enschede, The Netherlands), consisting of 17 IMUs, and a marker-based motion analysis system (Qualisys, Göteborg, Sweden), consisting of 8 infrared Oqus cameras and 42 passive reflective markers, were synchronized and used simultaneously to measure full-body kinematics. In the following, these measurement systems are referred to as the IMC and optical motion capture (OMC) systems. GRF&Ms were measured using two force plates instrumented in the laboratory floor (AMTI, Watertown, MA, USA), one under each foot. In Experiment II, the IMC system was used in combination with wireless sEMG (Noraxon, Scottsdale, AZ, USA) to measure full-body kinematics and the bilateral muscle activity of trapezius descendens and erector spinae longissimus, respectively. In Experiment III, OMC and force plates were used to measure full-body kinematics and GRF&Ms, similar to Experiment I.

2.3. EXPERIMENTAL PROCEDURES

2.3.1. EXPERIMENT I

After attaching the IMUs and reflective markers on the subjects, their body dimensions were measured with a caliper and input into the accompanying software, Xsens MVN Analyze v.2018.0.0 (Xsens Technologies BV, Encshede, The Netherlands). Hereafter, a calibration sequence was performed for the IMC system, which involved the subjects standing in a neutral posture (N-pose) and walking a few steps forward and back to the starting position.

Three repetitions of six lifting and two transferring tasks were performed: 1) a symmetrical lifting task with 5, 10, 15 and 20 kg, which involved the subjects lifting a box from the ground to an upright standing position and back to the starting position,

2) an asymmetrical lifting task where boxes weighing 5 and 10 kg were lifted from the ground to a 0.8 m high shelf placed 0.2 m to the right of the subjects, 3) a two-handed transferring task where a box of 10 kg was transferred between two tables of 1 m in height placed 0.2 m to the left and right of the subjects, and 4) a one-handed transferring task with 5 kg, similar to the two-handed task.

2.3.2. EXPERIMENT II

In the field-based risk assessment studies, four consecutive repetitions of 50 different MMH tasks were performed in two supermarkets. The choice of MMH tasks were based on observations made in similar stores and conversations with the industry stakeholders, and could be subdivided into four overall categories, namely fruit and vegetables, bread, meat and dairy, and colonial, i.e. edible and inedible goods with long shelf lives. Multiple start and end positions were included in the analysis, as the merchandise could be stocked from different starting positions to several shelf heights with varying depth. The height and depth of the start and end positions were indicated with the numbers 1-4 (low to high or closest to farthest). The subjects were informed about the start and end position as well as where to place their hands, but were otherwise encouraged to handle the merchandise as they normally would. If the boxes had handles, they were asked to use them. If not, they were asked to place and keep their hands on either side of the merchandise at approximately 1/3 of the boxes length. These restrictions were imposed to facilitate the musculoskeletal modelling procedures (see section 2.4). The task characteristics are described in detail in Sup. table 1a and 1b as well as Paper II and III.

After informing the subjects about the procedures, their mass and height were measured with a scale and caliper, respectively, and maximal voluntary isometric contractions (MVICs) were performed for the two muscle groups, which was later used for sEMG normalization. Then two investigators followed the subjects in to the shopping area with the merchandise assembled in a transport cage with two shelves (Low/High) and the measurement equipment on a rolling table. When performing a measurement, the investigators positioned the transport cage next to the shelf with the subject standing with their left side to the cage and the rolling table positioned opposite to the subject. From here, the subjects lifted the merchandise to the appropriate shelf on their right side, where after one of the investigators returned the merchandise back to the starting position. This procedure was repeated four times, where after the next series of four lifts was performed. When all tasks in a specific food category had been completed, the transport cage and rolling table was moved to the next area of the store, where a calibration of the IMC system was performed before the next series of measurements was initiated. The experimental procedures are illustrated in Fig. 2.1 and further described and illustrated in Paper II and III.

Figure 2.1 – Illustration of the experimental procedures for Experiment II. The subject is standing in a neutral posture during the calibration of the inertial motion capture system prior to performing MMH tasks in the meat and dairy (top) and fruit and vegetables areas (bottom). The surface electromyography electrodes are hidden under the subject's clothes.

2.3.3. EXPERIMENT III

Four consecutive repetitions of 21 lifting conditions were performed with systematic variations of load mass (LM), asymmetry angle (AA), horizontal (HL) and vertical location (VL), which are specified in Table 2.3. The lifting factors were inspired by the multipliers used in *The Revised NIOSH Lifting Equation* [88] with similar definitions of distances and angles between the subjects and the lifted loads. However, the VL refers to the end position and not the starting position of the load in the present study and was defined as the vertical distance from the shelf to the floor.

During the AA, HL and LM conditions, the subjects were instructed to lift the box to an upright standing position with their hands slightly above waist height and then lower it down to the starting position. During the VL condition, they were instructed to lift the box to the appropriate shelf. The subjects were encouraged to lift in a controlled fashion, but were otherwise free to lift the box in the manner they preferred. For all conditions, the initial lifting height was 25 cm (distance from the hands to the ground), while the horizontal locations were approximately 35 cm for the AA and VL conditions, and 45 cm for the LM condition. The experimental setup is illustrated in Fig. 2.2, while the procedures for each lifting condition are illustrated in Paper IV.

Table 2.3 - Overview of the lifting conditions in Experiment III with the abbreviation for each condition in cursive.

Load mass		**Asymmetry angle**		**Horizontal location**		**Vertical location**	
LM-5	5 kg	*AA-15*	15°	*HL-30*	30 cm	*VL-30*	30 cm
LM-10	10 kg	*AA-30*	30°	*HL-35*	35 cm	*VL-60*	60 cm
LM-15	15 kg	*AA-45*	45°	*HL-40*	40 cm	*VL-90*	90 cm
LM-20	20 kg	*AA-60*	60°	*HL-45*	45 cm	*VL-120*	120 cm
LM-25	25 kg	*AA-75*	75°	*HL-50*	50 cm	*VL-150*	150 cm
				HL-55	55 cm		
				HL-60	60 cm		

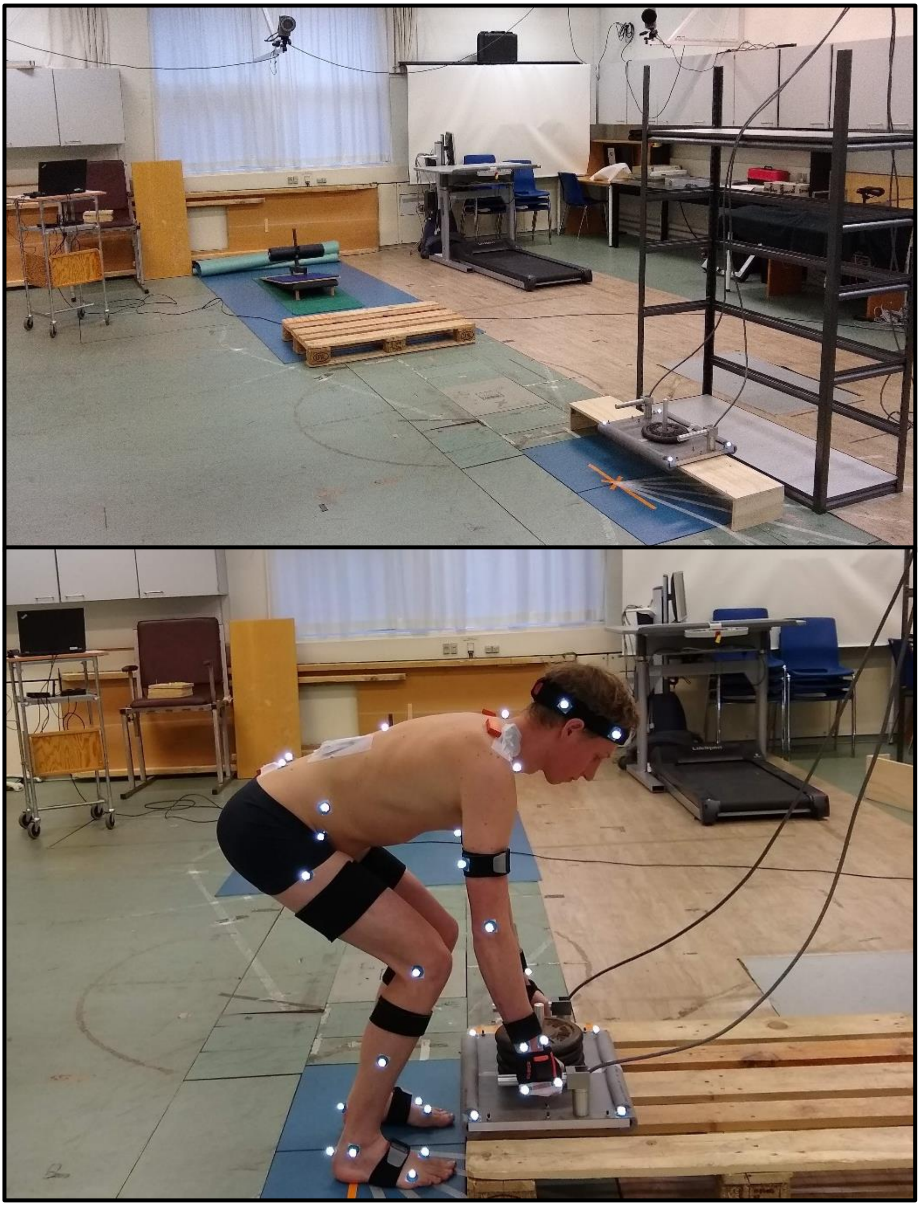

Figure 2.2 – Illustration of the experimental setup in Experiment III (top) and the initiation of the lift with a load mass of 25 kg (bottom).

2.4. MUSCULOSKELETAL MODELLING

Musculoskeletal models were developed based on the collected data from all three experiments. In Experiment I, three models were developed: one based on OMC with measured GRF&Ms (OMC-MGRF), one based on OMC with predicted GRF&Ms (OMC-PGRF) and one based on IMC with predicted ground reaction forces (IMC-PGRF). A flowchart illustrating the model development process can be found in Paper I. The method used for predicting GRF&Ms is described in section 2.4.2. The OMC-MGRF model was used as a silver standard, while the IMC-PGRF model was the configuration being evaluated for the field-based risk assessment (Experiment II). The OMC-PGRF model was included to evaluate the differences between models stemming from the different external force input and not the kinematic data. The models were developed in the AMS v. 7.1 using the Plug-in-gait-Multitrial_StandingRef (OMC-MGRF) and BVH_Xsens templates (IMC-PGRF) from the AnyBody Managed Model Repository v. 2.1 (AMMR). The model templates were identical except for how the kinematic data were handled in the AMS.

In Experiment II, the IMC-PGRF model was used to estimate joint loads based on IMC data obtained in two supermarkets. These models were developed in a later version of the AMS (v. 7.2) and AMMR (v. 2.2.3), but were otherwise identical.

In Experiment III, the OMC-MGRF model was used to estimate joint loads during standardized lifting activities in a laboratory setting. However, these models were also developed in a later version of the AMS (v. 7.3) and AMMR (v. 2.3). Two differences between the 7.2 and 7.3 versions that may have slightly affected the results were the introduction of a new experimental wrapping algorithm and additional muscle elements in the shoulders. Specifically, the pectoralis major was split into 10 muscle elements instead of five. Further details can be found in the AMMR documentation [140].

Common for all these versions were the base models used for the different body parts, as for instance, the lumbar spine, lower extremity and shoulder and arm models. The number of muscle elements in the models vary slightly between the different versions of the AMMR, which are specified in the appended papers. The lumbar spine model was based on the work of Hansen et al. [166], de Zee et al. [167] and Han et al. [168]. It consist of seven rigid segments, namely the lumbar vertebrae, the thoracic spine and sacrum, and is actuated by 188 muscle elements with representations of seven spinal ligaments and intra-abdominal pressure, similar to Han et al. [168]. The shoulder and arm model was based on the work of van der Helm et al. [137] and Veeger et al. [138,139], and includes 146 muscle elements, some of which wrap over analytical geometric shapes to mimic their complex wrapping behavior. Finally, the lower extremity model was based on the cadaver study of Carbone et al. [136] as well as the study of De Pieri et al. [169], and includes 169 muscle elements in each leg. The knee was modelled as a hinge joint with a fixed rotation center and axis with the patella tendon defined as a non-deformable element connecting the patella to the tibia.

For all experiments, computer-aided design models of the various boxes were developed in SolidWorks (Dassault Systems, Vélizy-Villacoublay Cedex, France). The mass and geometry of the boxes were based on measurements made during the experiments. This information was then used to estimate the inertial properties in SolidWorks.

2.4.1. MODEL SCALING AND KINEMATICS

Two different approaches were used for model scaling and kinematic analysis related to the use of two different types of kinematic input data. For the IMC-PGRF model used in Experiment I and II, the musculoskeletal models were scaled according to manually measured segment dimensions. The segment dimensions were input to the Xsens software prior to performing measurements and initially used to scale a 23 segment kinematic model (stick figure). After processing the kinematic data using the embedded tool in the software (HD-reprocess), Biovision Hierarchy files were exported, which contain a description of the kinematic model, the absolute position and orientation of the root pelvis segment as well as the joint angles between segments at each time frame [147]. To enable scaling and marker tracking of the musculoskeletal model based on the exported stick figure, the framework presented in Skals et al. [170] and Karatsidis et al. [147] was used. In short, the musculoskeletal models were mostly scaled according to the joint-to-joint distances of the stick figure, where after virtual markers were introduced on both the stick figure and musculoskeletal model to enable marker tracking (see Skals et al. [170] and Karatsidis et al. [147] for further details). For the OMC-MGRF model used in Experiment I and III, a single trial for each subject was initially used to determine segment lengths and marker positions using the optimization method of Andersen et al. [171]. The scaled segment lengths and marker positions were then saved and used to scale all other trials for that subject. For both the IMC-PGRF and OMC-MGRF model, the geometric and inertial parameters were scaled by applying a length-mass-fat scaling law [172] and the total body mass distributed to the body segments using the regression equations presented in Winter et al. [173]. Finally, the kinematics were solved by minimizing the least-square difference between model and experimental markers [174].

The kinematics of the lifted boxes were solved as follows: in Experiment I, rigid joints were defined between the hands and box for the IMC-PGRF model, so the box would follow the movement of the hands, while reflective markers were used to determine the box kinematics for the OMC-MGRF model. This discrepancy between modelling procedures potentially had some effect on the estimated forces, but was deemed acceptable, as the boxes were kept relatively stable during the execution of the lifts. In Experiment II, spherical joints were defined between the hands and boxes, meaning that the movement of the hands mostly determined the box translation and rotation.

However, an additional kinematic constraint was added to control the rotation of the boxes in the sagittal plane, which involved adding a point at the proximal and distal end on the right side, which had to remain at the same height relative to the ground. In Experiment III, the box kinematics were driven by the trajectories of reflective markers, similar to Experiment I.

2.4.2. PREDICTION OF EXTERNAL FORCES

In Experiment I and II, the IMC-PGRF model applied the method for predicting GRF&Ms first presented in Fluit et al. [142] and further developed and evaluated in Skals et al. [143] and Karatsidis et al. [144]. Twenty-five dynamic contact elements were attached under each foot of the musculoskeletal model. Each contact element consists of five uniaxial force actuators that were able to generate a positive normal force as well as positive and negative A-P and M-L static friction forces. A non-linear strength function was used to ensure that the contact elements would only generate forces when they were close to the ground and almost stationary, similar to Skals et al. [143]. Twelve contact elements with a high strength were also defined between the hands and boxes to estimate the external forces and moments. To improve numerical stability, small residual forces and moments were placed at the pelvis.

2.4.3. MUSCLE RECRUITMENT

For all models, the muscle, joint, contact and residual forces were distributed by solving a second (Experiment II and III) or third-order (Experiment I) optimization problem, which is commonly referred to as muscle recruitment [127]. In short, the optimization problem minimizes the muscle activities, or normalized muscle forces, and is constrained by the dynamic equilibrium equations, meaning that the solution must balance the external forces. In general, the higher the order of the optimization problem, the more muscles will be recruited to share the load. In all experiments, the muscles were modelled without contraction dynamics. The muscle strengths were determined from the physiological cross-sectional area and length-mass-fat scaling law [172,175]. A more detailed description of the second and third-order optimization problems can be found in Paper III-IV and I, respectively.

2.5. DATA ANALYSIS

The definition of the lifting cycles were generally similar across all three experiments. For nearly all lifts, the start and end points were defined as the instant when the subjects lifted the box from its base and the instant when the box made contact with the base at the end position, whether the end position was a table, shelf or pallet (see section 2.3). For the symmetrical lifts in Experiment I, the end point was defined as the time of maximal trunk extension, i.e. when the subjects were standing fully upright. The kinematic and kinetic data were resampled to 101 data points (one lifting cycle). All forces were normalized to percentage of bodyweight (%BW), while joint moments were normalized to body weight times body height (%BW x BH).

2.5.1. EXPERIMENT I

The following variables were extracted from the musculoskeletal models: trunk flexion, lateral bending and rotation angles, vertical GRF for the left and right foot, L4-L5 axial compression (A-C), A-P shear and M-L shear force.

2.5.2. EXPERIMENT II

The raw sEMG-signals were digitally filtered using a zero-phase, Butterworth fourth-order high-pass filter with a cut-off frequency of 10 Hz and a 500 ms moving root-mean-square filter. All raw and filtered signals were visually inspected to identify any signal quality issues and to assess if the filters had successfully removed noise and artefacts (see Paper II for further details). For each MMH task, the peak root-mean-square sEMG amplitudes were calculated for the four muscles and normalized to the absolute maximum sEMG amplitude of the MVICs (nEMG).

From the sEMG and kinematic data, the following variables were selected for further analysis: peak, 90th and 50th percentile nEMG for the left and right trapezius and erector spinae, trunk forward flexion (T8 relative to pelvis), lateral bending and rotation peak angles and range-of-motion, and bilateral knee and shoulder flexion peak angles and range-of-motion. From the musculoskeletal models, the peak L5-S1 A-C, A-P shear and M-L shear forces as well as the peak resultant JRF in the left and right knee and shoulder (glenohumeral) joints were extracted. Furthermore, the peak L5-S1 A-C and A-P shear forces were compared with the compression and shear tolerance limits of 3400 [54,88] and 1000 N [84,89], respectively, to assess the risk of injury to the lower back.

2.5.3. EXPERIMENT III

Similar to Experiment II, the peak L5-S1 A-C, A-P shear and M-L shear forces as well as the peak resultant JRF in the left and right knee and shoulder joints were extracted from the musculoskeletal models.

2.6. STATISTICAL ANALYSIS

To evaluate the differences between the IMC-PGRF and OMC-MGRF models in Experiment I, intraclass correlation coefficients (ICC), absolute (RMSE) and relative root-mean-square errors (rRMSE) were calculated for the time-series curves of each outcome variable. The ICCs were categorized as poor, moderate, good and excellent for ICC $\leq$ 0.5, 0.5 $\leq$ ICC $\leq$ 0.75, 0.75 $\leq$ ICC $\leq$ 0.9 and 0.9 < ICC, respectively

[176,177]. For this experiment, the statistical analyses were performed in MATLAB R2018b (MathWorks Inc., Natick, MA, USA).

For the risk assessment studies (Experiment II), the main purpose of the statistical analyses was to determine least square means with 95% confidence intervals for each MMH task in order to rank the tasks from highest to lowest for each outcome variable, e.g. the L5-S1 JRFs, trunk kinematics and nEMG. Repeated measures linear mixed models (Proc Mixed, SAS) were used for this purpose with the forces, joint angles and muscle activities as the dependent variables, and the MMH tasks treated as a single variable and included as a fixed effect. To check whether the model assumptions were met, residual diagnostics plots were inspected to ensure a normal distribution of the residuals and homogeneity of variance. Within subject correlation was assumed and modelled as a random effect. The covariance structure was set to Variance Components and the model fit using restricted maximum likelihood estimation. The confidence intervals were based on a Satterwaite approximation. The statistical analyses were performed in SAS v. 9.4 (SAS Institute Inc., Cary, NC, USA).

In Experiment III, a similar approach as in Experiment II was used to test the statistical differences between the condition levels. Specifically, repeated measures linear mixed models were used to test if any significant differences existed between the different levels for each condition separately. The peak JRFs were the dependent variables, while the condition levels were included as fixed effects. However, for these analyses, differences of least square means were also presented in addition to the least square means with 95% confidence intervals. Significant differences were reported for $p < 0.05$.

CHAPTER 3. RESULTS

In the following, a summary of the main results from Paper I-IV are presented. Time-series curves and tables for all outcome variables related to Paper I and IV can be found in the appended papers. For Paper II and III, the full dataset can be found in an online supplementary database [178], which has also been included in Appendix A.

3.1. PAPER I

The trunk kinematics showed poor agreement between the IMC-PGRF and OMC-MGRF models. Specifically, the trunk forward flexion angle showed poor ICCs for all analyzed tasks (from 0.20 to 0.41) as well as notable RMSEs (from 6.9 to 16°) and rRMSEs (from 129 to 247%). Similarly, trunk lateral bending showed poor ICCs for all analyzed tasks (from -0.01 to 0.41), but low RMSEs (from 1.8 to 2.9°), as the lateral bending angles were generally very low. For trunk rotation, the generally low values also contributed to poor ICCs for the symmetrical and asymmetrical lifts (ICC: 0.01-0.24, RMSE: 4.6-5.4°) with the exception of the one and two-handed transferring tasks, which showed substantially higher rotation angles and good ICCs (0.79 and 0.83). There were, however, still some magnitude differences for trunk rotation during the transferring tasks (RMSE: 4.6 and 7.8°, rRMSE: 19 and 16%).

For the kinetic variables, the agreement between model estimates was generally better in comparison. The L4-L5 A-C force showed good to excellent ICCs for the symmetrical and asymmetrical lifting tasks (from 0.85 to 0.92) and reasonable magnitude differences (RMSE: 50-75 %BW, rRMSE: 23-34%). For the one and two-handed transferring tasks, the ICCs were moderate (0.57) and poor (0.16) with RMSEs of 64 and 45 %BW, respectively. The L4-L5 A-P shear force showed moderate to good ICCs across all the analyzed tasks (from 0.65 to 0.79), but more notable magnitude differences (RMSE: 8.0-23 %BW, rRMSE: 35-58%). The L4-L5 M-L shear force showed poor to moderate ICCs across the analyzed tasks (from 0.01 to 0.51), but less severe magnitude differences, specifically RMSEs ranging from 1.7 to 4.1 %BW and rRMSEs from 50 to 127%. Finally, the vertical GRFs showed moderate to excellent ICCs (from 0.51 to 0.96) and low magnitude differences in general with RMSEs ranging from 5.1 to 12.1 %BW and rRMSEs from 10.3 to 34.1%. An excerpt of the results is listed in Table 3.1.

Table 3.1 - Intraclass correlation coefficient (ICC), root-mean-square error (RMSE) and relative RMSE (rRMSE) for the trunk kinematics and L4-L5 axial compression (A-C), anteroposterior (A-P) shear and mediolateral (M-L) shear forces during symmetrical lifting with 10 (SYM-10) and 20 kg (SYM-20), asymmetrical lifting with 10 kg (ASYM-10), one (TRA-OH) and two-handed transferring (TRA-BOX). The RMSE and rRMSE are presented as the mean ± SD. The table is adapted from Paper I and contains an excerpt of the results.

	SYM-10	**SYM-20**	**ASYM-10**	**TRA-BOX**	**TRA-OH**
			ICC		
Trunk flexion	0.41	0.31	0.28	0.20	0.21
Trunk lateral bending	0.16	0.10	0.19	0.28	0.41
Trunk rotation	0.09	0.01	0.47	0.79	0.83
A-C force	0.90	0.85	0.92	0.57	0.16
A-P shear force	0.77	0.73	0.79	0.78	0.74
M-L shear force	0.43	0.09	0.51	0.23	0.04
			RMSE		
Trunk flexion	15 ± 7	17 ± 8	13 ± 7	6.9 ± 4.4	7.3 ± 6.5
Trunk lateral bending	2.3 ± 1.0	2.1 ± 1.2	2.8 ± 1.5	2.5 ± 1.4	2.9 ± 1.5
Trunk rotation	4.8 ± 3.4	5.0 ± 3.0	5.4 ± 4.0	7.8 ± 3.3	5.1 ± 3.2
A-C force	67 ± 34	75 ± 57	57 ± 31	45 ± 24	64 ± 39
A-P shear force	19 ± 11	23 ± 16	15 ± 8	8.4 ± 4.3	8.0 ± 5.2
M-L shear force	2.4 ± 1.5	2.7 ± 1.6	4.0 ± 1.8	8.4 ± 4.3	8.0 ± 5.2
			rRMSE		
Trunk flexion	170 ± 234	198 ± 183	187 ± 208	247 ± 219	203 ± 175
Trunk lateral bending	181 ± 132	130 ± 76	82 ± 76	31 ± 21	41 ± 66
Trunk rotation	165 ± 142	229 ± 211	36 ± 27	19 ± 7	16 ± 11
A-C force	29 ± 12	28 ± 21	25 ± 15	67 ± 34	113 ± 61
A-P shear force	46 ± 29	50 ± 38	37 ± 29	57 ± 38	58 ± 38
M-L shear force	79 ± 38	97 ± 66	63 ± 27	50 ± 34	62 ± 72

3.2. PAPER II

Of the 17 subjects who participated in the study, 15 were included in the final analysis. From these subjects, 2922 IMC trials as well as 2672, 2774, 2611 and 2727 trials of muscle activity data for the left and right trapezius descendens and erector spinae longissimus, respectively, were included in the analysis. The exclusion of subjects and trials were described in detail in Paper II.

The linear mixed model analyses showed significant differences for all outcome variables ($p < 0.001$). The least square means with 95% confidence intervals, where the MMH tasks are ranked from highest to lowest for each outcome, and 50 figures illustrating the joint angles over the complete lifting cycles can be found in Appendix A as well as the supplementary database [178]. However, excerpts of the results for the peak trapezius and erector spinae muscle activities, peak knee and shoulder flexion as well as peak trunk flexion and rotation angles are listed in Tables 3.2-3.6. The muscle activities and joint angles are presented as percentage of MVIC and in degrees, respectively.

For the bilateral trapezius muscle activity, the highest ranked tasks were Bread-HighToHigh (59 and 56%), Bread-LowToHigh (55 and 56%), Cucumbers-HighToHigh (53 and 50%), ColdCuts-HighToHighFar (53 and 50%) and Cucumbers-LowToHigh (51 and 47%). As can be seen in Table 3.2, the 10 highest ranked tasks were all variations of bread, cucumbers, cold cuts and yoghurts lifted to the highest shelf heights (108-168 cm above floor level). In general, the highest muscle activities were found when the relatively heavy merchandise, e.g. bread (7.9 kg) and cucumbers (10.2 kg), were lifted to the highest shelf heights. The trapezius muscle activities ranged from 3 to 59% with a median of 22% across all analyzed tasks (see Sup. table 2a and 2b).

For the bilateral erector spinae muscle activity, the highest ranked tasks were Cucumbers-LowToHigh (67 and 71%), Bread-LowToHigh (60 and 63%), Bananas-LowToLow (59 and 63%) and Milk-LowToHigh (61 and 61%). As can be seen in Table 3.3., the highest muscular efforts in the lower back were found when the relatively heavy merchandise was lifted from a low position (15 cm above floor level), namely bananas (20.2 kg), milk crates (17.3 kg), cucumbers and bread. Across all the 50 analyzed tasks, the erector spinae muscle activities ranged from 18 to 71% with a median of 43% (see Sup. table 2a and 2b).

The MMH tasks showing the highest amount of knee flexion were Yoghurts-LowToHigh (108° and 108°), Yoghurts-LowToLow (105° and 107°), ColdCuts-HighToLowFar (98° and 100°) and ColdCuts-LowToHighNear (94° and 98°). The distinctive features of these tasks were that they involved smaller, narrow boxes either lifted to or from a low starting position. The top 10 highest ranked tasks all required flexing the knees 89° or more (see Table 3.2).

The tasks requiring the highest amount of shoulder flexion were ColdCuts-HighToLowFar (109° and 110°), Bread-LowToLow (105° and 106°), Herbs-HighToHigh (102° and 109°) and Bread-LowToHigh (102° and 107°). Almost all the 25 highest ranked tasks required flexing the shoulders nearly 90° and typically involved placing merchandise at either the lowest or the highest shelves (see Sup. table 5a and 5b).

For the peak trunk flexion angle, variations of handling tomato cans and cold cuts generally showed the highest values, e.g. TomatoCans-LowToLow (59°) and ColdCuts-LowToHighNear (56°). Twenty-two of the analyzed tasks required flexing the trunk 50° or more. All these tasks involved lifting from or to a low position. For trunk rotation, the highest ranked tasks were VegetableOil-HighToLow (24°), ColdCuts-HighToLowFar (24°) and VegetableOil-HighToHigh (22°). Several of the one-handed tasks showed relatively high degree of trunk rotation. Across all 50 analyzed tasks, peak trunk rotation ranged from 9 to 24° with a median of 17°.

Table 3.2 - Peak muscle activities for the left (L) and right (R) trapezius descendens presented as percentage of maximal voluntary isometric contraction (%MVIC) with 95% confidence intervals for the 10 highest ranked manual material handling tasks. The table is adapted from Paper II and contains an excerpt of the results.

	Trapezius descendens (L)		**Trapezius descendens (R)**	
Rank	**Task**	**%MVIC**	**Task**	**%MVIC**
1	Bread-HighToHigh	59 (54 – 64)	Bread-LowToHigh	56 (51 – 62)
2	Bread-LowToHigh	55 (50 – 61)	Bread-HighToHigh	56 (51 – 62)
3	Cucumbers-HighToHigh	53 (47 – 58)	Cucumbers-LowToHigh	51 (46 – 56)
4	ColdCuts-HighToHighFar	53 (47 – 58)	Cucumbers-HighToHigh	50 (44 – 55)
5	Cucumbers-LowToHigh	47 (42 – 53)	ColdCuts-HighToHighFar	50 (44 – 55)
6	Bread-v2-HighToHigh	44 (39 – 50)	Bread-v2-HighToHigh	47 (41 – 52)
7	Yoghurts-HighToHigh	42 (37 – 48)	Yoghurts-HighToHigh	45 (39 – 50)
8	Yoghurts-LowToHigh	40 (34 – 45)	Yoghurts-LowToHigh	42 (37 – 47)
9	ColdCuts-HighToHighNear	32 (27 – 38)	ColdCuts-HighToHighNear	41 (36 – 47)
10	ColdCuts-LowToHighNear	31 (25 – 36)	ColdCuts-LowToHighNear	35 (30 – 41)

Table 3.3 - Peak muscle activities for the left (L) and right (R) erector spinae longissimus presented as percentage of maximal voluntary isometric contraction (%MVIC) with 95% confidence intervals for the 10 highest ranked manual material handling tasks. The table is adapted from Paper II and contains an excerpt of the results.

	Erector spinae longissimus (L)		**Erector spinae longissimus (R)**	
Rank	**Task**	**%MVIC**	**Task**	**%MVIC**
1	Cucumbers-LowToHigh	67 (58 – 75)	Cucumbers-LowToHigh	71 (62 – 80)
2	Milk-LowToHigh	61 (52 – 69)	Cucumbers-HighToHigh	64 (55 – 72)
3	Bread-LowToHigh	60 (51 – 68)	Bread-LowToHigh	63 (54 – 71)
4	Bananas-LowToLow	59 (51 – 68)	Bananas-LowToLow	63 (54 – 71)
5	Cucumbers-HighToHigh	57 (48 – 65)	Cucumbers-LowToMid	62 (53 – 70)
6	Cucumbers-LowToMid	56 (48 – 65)	Milk-LowToHigh	61 (53 – 70)
7	Bread-HighToHigh	56 (47 – 65)	Salads-LowToHigh	59 (51 – 68)
8	Bread-LowToMid	55 (46 – 63)	Yoghurts-LowToHigh	58 (50 – 67)
9	Milk-LowToMid	53 (44 – 61)	Milk-LowToMid	57 (48 – 65)
10	Yoghurts-LowToHigh	51 (43 – 60)	Bread-HighToHigh	56 (48 – 64)

Table 3.4 - Peak knee flexion angles for the left (L) and right (R) side presented in degrees with 95% confidence intervals for the 10 highest ranked manual material handling tasks. The table is adapted from Paper II and contains an excerpt of the results.

	Knee flexion angle (L)		**Knee flexion angle (R)**	
Rank	**Task**	**Peak**	**Task**	**Peak**
1	Yoghurts-LowToHigh	108 (100 – 116)	Yoghurts-LowToHigh	108 (99 – 117)
2	Yoghurts-LowToLow	105 (97 – 113)	Yoghurts-LowToLow	107 (98 – 116)
3	ColdCuts-HighToLowFar	98 (90 – 106)	ColdCuts-HighToLowFar	100 (91 – 109)
4	ColdCuts-LowToHighNear	94 (86 – 102)	ColdCuts-LowToHighNear	98 (89 – 107)
5	Cucumbers-LowToMid	94 (85 – 102)	TomatoCans-LowToMid	94 (85 – 103)
6	Cucumbers-LowToHigh	91 (83 – 99)	ColdCuts-LowToMidNear	93 (84 – 103)
7	TomatoCans-LowToMid	90 (82 – 98)	Cucumbers-LowToHigh	93 (84 – 102)
8	ColdCuts-LowToMidNear	90 (82 – 98)	Cucumbers-LowToMid	93 (83 – 102)
9	Salads-LowToHigh	90 (82 – 98)	TomatoCans-LowToLow	90 (81 – 99)
10	Herbs-LowToHigh	89 (81 – 97)	ColdCuts-LowToLowNear	89 (80 – 98)

Table 3.5 - Peak shoulder flexion angles for the left (L) and right (R) side presented in degrees with 95% confidence intervals for the 10 highest ranked manual material handling tasks. The table is adapted from Paper II and contains an excerpt of the results.

	Shoulder flexion angle (L)		**Shoulder flexion angle (R)**	
Rank	**Task**	**Peak**	**Task**	**Peak**
1	ColdCuts-HighToLowFar	109 (104 – 113)	ColdCuts-HighToLowFar	110 (105 – 116)
2	Bread-LowToLow	105 (101 – 110)	SingleYoghurt-HighToHigh	109 (104 – 115)
3	Herbs-HighToHigh	102 (98 – 107)	Herbs-HighToHigh	109 (103 – 114)
4	Bread-LowToHigh	102 (98 – 107)	Bread-LowToHigh	107 (102 – 113)
5	Bread-HighToHigh	102 (98 – 107)	Herbs-LowToHigh	107 (101 – 113)
6	Herbs-LowToHigh	101 (97 – 105)	Bread-LowToLow	106 (100 – 112)
7	Bread-HighToLow	100 (95 – 104)	VegetableOil-HighToHigh	105 (99 – 111)
8	ColdCuts-LowToLowNear	97 (93 – 101)	Bread-HighToHigh	104 (99 – 110)
9	ColdCuts-HighToHighFar	96 (92 – 100)	ColdCuts-HighToHighFar	104 (98 – 109)
10	ColdCuts-LowToHighNear	96 (92 – 100)	Salads-LowToHigh	103 (97 – 108)

Table 3.6 - Peak trunk flexion and rotation angles presented in degrees with 95% confidence intervals for the 10 highest ranked manual material handling tasks. The table is adapted from Paper II and contains an excerpt of the results.

	Trunk flexion angle		**Trunk rotation angle**	
Rank	**Task**	**Peak**	**Task**	**Peak**
1	TomatoCans-LowToLow	59 (54 – 63)	VegetableOil-HighToLow	24 (20 – 27)
2	Bread-LowToLow	58 (53 – 62)	ColdCuts-HighToLowFar	24 (20 – 27)
3	ColdCuts-LowToHighNear	56 (52 – 61)	VegetableOil-HighToHigh	22 (19 – 25)
4	ColdCuts-LowToLowNear	56 (52 – 61)	ColdCuts- HighToMidFar	22 (19 – 25)
5	ColdCuts-HighToLowFar	56 (52 – 61)	Bread-v2-HighToMid	22 (19 – 25)
6	TomatoCans-LowToMid	56 (52 – 61)	Bread-HighToLow	21 (18 – 25)
7	TomatoCans-HighToMid	56 (51 – 60)	MincedBeef-HighToLowFar	21 (18 – 25)
8	ColdCuts-LowToMidNear	56 (51 – 60)	VegetableOil-HighToMid	21 (18 – 24)
9	Cucumbers-LowToHigh	56 (51 – 60)	Herbs-HighToHigh	21 (18 – 24)
10	Cucumbers-LowToMid	55 (51 – 60)	MincedBeef-HighToLowNear	21 (18 – 24)

3.3. PAPER III

The 2922 IMC trials were initially used to drive the musculoskeletal models. However, due to several issues with the modelling procedures, a large number of trials were excluded. These issues were mostly related to inaccuracies of the hand positions in the kinematic data and errors in the muscle wrapping of the wrist flexors (see Paper III for further details). Because of these issues, all tasks involving smaller, narrow boxes as well as all one-handed lifts were excluded from this part of the analysis. However, 1479 trials of the 26 relatively heavy, two-handed tasks were successfully modelled and included in the final analysis (see Sup. table 9). Musculoskeletal models of three of the included tasks are illustrated in Fig. 2.3 and 2.4.

The linear mixed model analyses showed significant differences for the fixed effect (MMH tasks) for each outcome variable ($p < 0.0001$). Least square means with 95% confidence intervals of the L5-S1, knee and shoulder JRFs for all 26 MMH tasks are listed in Sup. tables 10, 11 and 12, while an excerpt of the results are presented in Tables 3.7-3.9. Time-series curves of the JRFs are illustrated in Sup. figures 51-76. Similar to Paper II, the MMH tasks are ranked from highest to lowest for each outcome variable.

The handling of bananas (553 and 539 %BW) and milk crates (from 424 to 506 %BW) resulted in the highest L5-S1 A-C forces. Cucumbers lifted from the low starting position (from 413 to 449 %BW) as well as bread placed on the lowest shelf (442 and 425 %BW) also showed relatively high A-C forces. Similar results were found for the L5-S1 A-P shear force with the handling of bananas (142 and 155 %BW) and milk lifted from or to a low position (from 134 to 144 %BW) showing the highest forces, followed by cucumbers and bread. The L5-S1 M-L shear forces were generally low (from 5 to 16 %BW) and quite similar across the analyzed tasks.

There were 8 and 6 tasks where the upper confidence limit for the L5-S1 A-C and A-P shear forces exceeded the biomechanical tolerance limits of 3400 and 1000 N, respectively (see Table 3.7). For example, Bananas-LowToLow (4188 N for A-C and 1191 N for A-P shear force), Bananas-HighToLow (4088 and 1097 N) and Milk-LowToLow (3854 and 1113 N).

The highest knee resultant JRFs were found during the handling of bananas (761 and 848 %BW), milk crates (from 637 to 799 %BW) and cucumbers (from 625 to 751 %BW). The results were slightly different for the left and right leg with the handling of bananas showing the highest forces in the left knee, while handling milk resulted in the highest forces in the right knee.

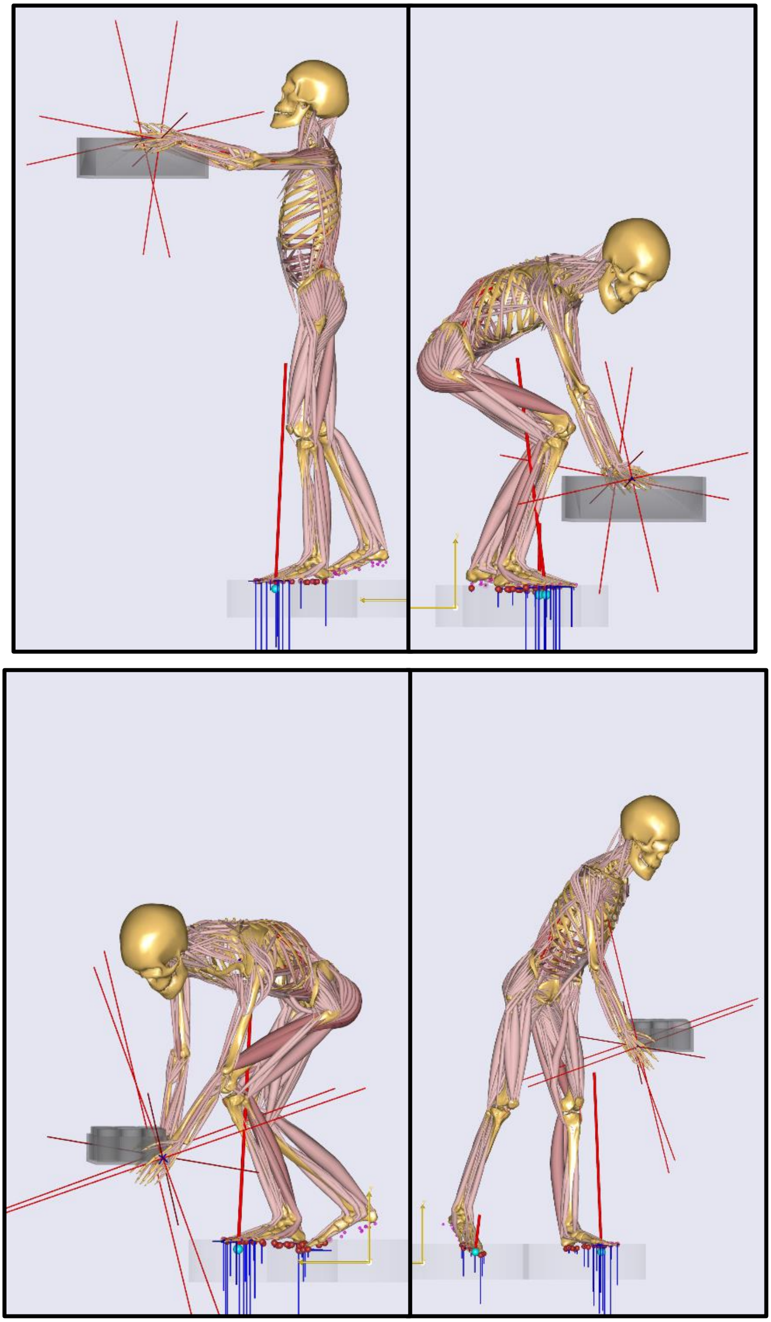

Figure 2.3 – Musculoskeletal models of the task Bread-LowToHigh (top) and TomatoCans-HighToLow (bottom) at the start (right) and end (left) of the lifting cycle.

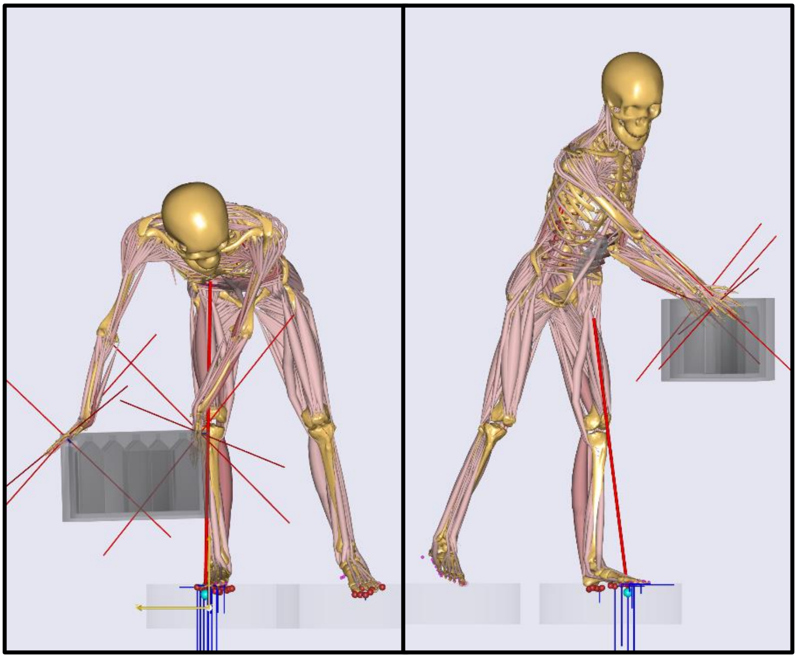

Figure 2.4 - Musculoskeletal model of the task Milk-HighToLow at the start (right) and end (left) of the lifting cycle.

Table 3.7 - Peak axial compression and anteroposterior shear forces presented in percentage of bodyweight (%BW) with 95% confidence intervals for the 10 highest ranked manual material handling tasks. The table is adapted from Paper III and contains an excerpt of the results.

	L5-S1 axial compression force		**L5-S1 anteroposterior force**	
Rank	**Task**	**%BW**	**Task**	**%BW**
1	Bananas-LowToLow	553 (530 – 576)	Bananas-LowToLow	155 (147 – 162)
2	Bananas-HighToLow	539 (516 – 562)	Milk-LowToLow	144 (137 – 152)
3	Milk-LowToLow	506 (482 – 529)	Bananas-HighToLow	142 (134 – 149)
4	Milk-LowToMid	501 (478 – 524)	Milk-LowToMid	139 (131 – 147)
5	Milk-LowToHigh	494 (471 – 517)	Milk-LowToHigh	137 (130 – 145)
6	Milk-HighToLow	489 (466 – 512)	Milk-HighToLow	134 (126 – 141)
7	Milk-HighToMid	467 (443 – 490)	Cucumbers-LowToHigh	126 (118 – 134)
8	Cucumbers-LowToHigh	449 (426 – 472)	Cucumbers-LowToMid	124 (116 – 132)
9	Cucumbers-LowToMid	442 (418 – 465)	Bread-LowToLow	119 (111 – 126)
10	Bread-LowToLow	425 (401 – 448)	Bread-HighToLow	118 (110 – 126)

Table 3.8 - Peak glenohumeral joint reaction forces (JRF) for the left (L) and right (R) side presented in percentage of bodyweight (%BW) with 95% confidence intervals for the 10 highest ranked manual material handling tasks. The table is adapted from Paper III and contains an excerpt of the results.

	Glenohumeral resultant JRF (L)		**Glenohumeral resultant JRF (R)**	
Rank	**Task**	**%BW**	**Task**	**%BW**
1	Cucumbers-LowToHigh	225 (212 – 237)	Cucumbers-LowToHigh	227 (214 – 240)
2	Bread-HighToHigh	223 (211 – 235)	Bread-HighToHigh	225 (212 – 238)
3	Cucumbers-HighToHigh	222 (210 – 235)	Cucumbers-HighToHigh	220 (207 – 233)
4	Bread-LowToHigh	217 (205 – 230)	Bread-LowToHigh	217 (204 – 230)
5	Bananas-HighToLow	167 (155 – 179)	Bananas-HighToLow	160 (147 – 173)
6	Bananas-LowToLow	140 (128 – 153)	Bananas-LowToLow	144 (131 – 157)
7	Cucumbers-HighToMid	135 (122 – 148)	Cucumbers-LowToMid	137 (123 – 150)
8	Cucumbers-LowToMid	133 (121 – 146)	Cucumbers-HighToMid	135 (121 – 148)
9	Salads-HighToHigh	124 (112 – 137)	Milk-HighToHigh	132 (119 – 145)
10	Salads-LowToHigh	121 (108 – 133)	Milk-LowToHigh	130 (117 – 143)

Table 3.9 - Peak knee resultant joint reaction forces (JRF) for the left (L) and right (R) side presented in percentage of bodyweight (%BW) with 95% confidence intervals for the 10 highest ranked manual material handling tasks. The table is adapted from Paper III and contains an excerpt of the results.

	Knee resultant JRF (L)		**Knee resultant JRF (R)**	
Rank	**Task**	**%BW**	**Task**	**%BW**
1	Bananas-HighToLow	848 (797 – 900)	Milk-HighToHigh	799 (746 – 853)
2	Bananas-LowToLow	761 (709 – 813)	Milk-LowToMid	769 (715 – 824)
3	Cucumbers-HighToHigh	751 (698 – 804)	Milk-HighToMid	760 (706 – 814)
4	Cucumbers-LowToHigh	727 (674 – 780)	Milk-LowToLow	752 (698 – 806)
5	Cucumbers-HighToMid	695 (639 – 750)	Cucumbers-LowToHigh	750 (695 – 805)
6	Milk-HighToLow	684 (632 – 735)	Milk-HighToLow	747 (693 – 801)
7	Bread-HighToHigh	681 (629 – 733)	Milk-LowToHigh	744 (690 – 798)
8	Cucumbers-LowToMid	681 (625 – 736)	Bananas-LowToLow	735 (681 – 789)
9	Salads-HighToMid	671 (619 – 722)	Bananas-HighToLow	705 (651 – 759)
10	Milk-HighToMid	668 (617 – 720)	Cucumbers-HighToHigh	666 (611 – 721)

In general, lifting relatively heavy merchandise to high shelf heights resulted in the highest shoulder resultant JRFs. The highest ranked tasks for both the left and right shoulder involved lifting cucumbers (from 220 to 227 %BW) and bread (from 217 to 225 %BW) to the highest shelf heights. The next highest ranked tasks were the handling of bananas (from 140 to 167 %BW), but these forces were considerably lower.

3.4. PAPER IV

A total of 1832 lifting trials were included in the analysis. Least square means with 95% confidence intervals, indications of significant differences between condition levels as well as time-series curves for all lifting conditions can be found in Paper IV. In the following, the results are summarized and an excerpt of the results listed in Tables 3.10 and 3.11.

LM had a substantial influence on most of the joint forces with significant differences between nearly all condition levels across the analyzed variables. The L5-S1 A-C forces ranged from 472 to 672 %BW, while the A-P shear forces ranged from 48 to 67 %BW for the 5 to 25 kg increments in LM. The M-L force was less affected by the increased LM and ranged from 6 to 10 %BW. The left (from 425 to 530 %BW) and right knee resultant JRFs (from 453 to 537 %BW) also increased with approximately 100 %BW between the lowest and highest level. Finally, the left and right shoulder resultant JRFs increased from 40 to 128 %BW and 40 to 126 %BW, respectively.

The increments in AA had a negligible effect on the L5-S1 A-C force (from 513 to 543 %BW), but a more substantial influence on the A-P (from 52 to 68 %BW) and M-L shear forces (from 14 to 40 %BW). Furthermore, the AA also had a significant effect on the knee resultant JRFs. As the load was shifted towards the right side, the right knee forces (439 to 679 %BW) increased, while the left knee forces decreased (from 348 to 263 %BW). Finally, every increment in AA led to significantly higher left (from 63 to 107 %BW) and right shoulder JRFs (from 57 to 113 %BW).

The HLs had a minor effect on the L5-S1 A-C (from 502 to 549 %BW), A-P shear (from 50 to 55 %BW) and M-L shear forces (from 7 to 8 %BW), but a substantial influence on the knee and shoulder JRFs. Specifically, the JRFs in the left (from 362 to 466 %BW) and right knee (from 388 to 503 %BW) significantly increased with nearly every increment in HL. Similarly, the left (from 53 to 104 %BW) and right shoulder JRFs (from 53 to 102 %BW) significantly increased as a result of the increased HL. The largest effect on the shoulder JRFs was found when the HL was 45 cm or above.

Finally, the increments in VL resulted in a slight decrease in the L5-S1 A-C forces (from 519 to 498 %BW) and a slight increase in the A-P (from 45 to 54 %BW) and M-L shear forces (from 6 to 8 %BW). It should be noted, however, that the peak L5-

S1 A-C forces mostly occurred at the initiation of the lifts rather than when placing the box on the shelves. The VLs had a minor effect on the JRFs in the left (from 305 to 346 %BW) and right knee (from 321 to 380 %BW), but a substantial influence on the left (from 117 to 240 %BW) and right shoulders (from 113 to 245 %BW), which more than doubled from the lowest to the highest VL.

Table 3.10 - Peak L5-S1 axial compression (A-C), anteroposterior (A-P) shear and mediolateral (M-L) shear forces during the load mass (LM) and asymmetry angle (AA) conditions normalized to percentage of bodyweight. The table is adapted from Paper IV and contains an excerpt of the results.

	L5-S1 A-C	**L5-S1 A-P**	**L5-S1 M-L**
LM-5	472 (438 - 506)	47.5 (36.8 - 58.2)	6.2 (4.5 - 7.9)
LM-10	522 (489 - 556)	51.4 (40.7 - 62.1)	6.5 (4.8 - 8.2)
LM-15	569 (535 - 603)	56.8 (46.1 - 67.5)	7.6 (5.9 - 9.3)
LM-20	616 (582 - 650)	60.4 (49.8 - 71.1)	8.1 (6.3 - 9.8)
LM-25	672 (638 - 706)	67.0 (56.3 - 77.7)	9.5 (7.8 - 11.2)
AA-15	513 (476 - 550)	51.9 (41.5 - 62.2)	14.1 (10.1 - 18.1)
AA-30	530 (493 - 567)	56.0 (45.7 - 66.3)	22.6 (18.6 - 26.6)
AA-45	531 (495 - 568)	58.7 (48.4 - 69.0)	27.4 (23.4 - 31.5)
AA-60	531 (493 - 568)	62.5 (52.1 - 72.8)	33.2 (29.2 - 37.3)
AA-75	543 (506 - 580)	67.8 (57.4 - 78.1)	40.0 (36.0 - 44.1)

Table 3.11 - Peak knee and shoulder (glenohumeral) resultant joint reaction forces (JRFs) for the left (L) and right side (R) during the horizontal (HL) and vertical location (VL) conditions normalized to percentage of bodyweight. The table is adapted from Paper IV and contains an excerpt of the results.

	Knee JRF (L)	**Knee JRF (R)**	**Shoulder JRF (L)**	**Shoulder JRF (R)**
HL-30	362 (306 – 418)	388 (335 - 442)	53 (45 - 61)	53 (46 - 61)
HL-35	399 (343 - 455)	430 (376 - 483)	56 (48 - 64)	57 (49 - 64)
HL-40	427 (371 - 483)	446 (393 - 499)	59 (51 - 67)	60 (53 – 68)
HL-45	454 (398 - 511)	483 (430 - 537)	65 (57 - 73)	64 (57 - 72)
HL-50	435 (379 - 492)	467 (413 - 520)	71 (63 - 79)	72 (64 - 79)
HL-55	450 (394 - 506)	483 (430 - 536)	86 (78 - 94)	85 (78 - 93)
HL-60	466 (410 - 523)	503 (449 - 556)	104 (96 - 112)	102 (95 - 110)
VL-30	305 (273 - 336)	321 (285 - 357)	117 (97 - 136)	113 (93 - 132)
VL-60	330 (299 - 362)	354 (318 – 390)	133 (114 - 153)	131 (111 - 150)
VL-90	328 (297 - 360)	358 (322 - 395)	166 (146 - 186)	165 (146 - 183)
VL-120	363 (332 - 395)	393 (357 - 429)	201 (182 - 221)	201 (182 - 220)
VL-150	346 (314 - 377)	380 (344 - 417)	240 (221 - 260)	245 (226 - 265)

CHAPTER 4. DISCUSSION

This dissertation developed and evaluated a novel methodology for field-based analysis of MMH using state-of-the-art musculoskeletal models (Paper I), which was then applied in combination with sEMG for a comprehensive risk assessment in two supermarkets (Paper II and III). Similar models were also applied to study the relative importance of several lifting factors on multiple joint loads in a laboratory setting (Paper IV), hereby providing context for the field-based estimates as well as valuable reference data for advising on the regulation of work-related MMH. In the following, these findings are discussed in detail.

4.1. MAIN FINDINGS

There were several important findings of the dissertation. First, the evaluation of the methodology for field-based risk assessment of MMH showed that the A-C force in the lumbar spine could be estimated with reasonable accuracy during standardized lifting activities when compared to models driven by OMC and force plate data. Hence, we assessed that the methodology was sufficiently accurate to incorporate in the field-based risk assessment, as the A-C force was the primary outcome of interest. However, considerable discrepancies were also identified between the two methodologies, particularly with respect to the estimation of the trunk flexion angle and A-P shear force. The implications of these discrepancies are discussed in section 4.1.1. Second, based on the two risk assessment studies, we identified several MMH tasks that may pose a risk for developing low back pain and injury, as the estimated L5-S1 A-C and A-P shear forces exceeded well-known biomechanical tolerance limits and required relatively high muscular efforts in the lower back. We also found that lifting relatively heavy merchandise to the highest shelf heights led to considerably higher loads in the shoulders compared with the other tasks based on both the estimated shoulder JRFs and muscular efforts in the neck and shoulder region. Furthermore, it was evident from the kinematic data that the stocking work performed in the supermarkets involved many undesirable working postures, e.g. high degrees of forward bending, lifting to above shoulder height and excessive squatting during lifts. Finally, the novelty of the laboratory-based lifting analysis was the simultaneous estimation of multiple dynamic joint loads during variations of work-related lifting, which is very rare and has never been performed to this extent using musculoskeletal models of this level of complexity. From this study, we found that LM had the most considerable effect on the JRFs overall, particularly with respect to the L5-S1 JRFs, while increments in the AA led to considerably higher L5-S1 M-L shear forces, knee and shoulder resultant JRFs. Furthermore, the HL had a negligible influence on the spinal forces, but a more substantial influence on the knee and shoulder JRFs, while the VL had the most substantial influence on the shoulder forces, but a minor effect on the forces in the knees and lumbar spine.

4.1.1. EVALUATION OF METHODOLOGY

The main findings of Paper I were that the IMC-PGRF model showed moderate to excellent ICCs (from 0.51 to 0.96) and relatively low magnitude differences (rRMSEs ranging from 11 to 34%) for the L4-L5 A-C force and vertical GRFs during symmetrical and asymmetrical lifting with different loads. Based on these results, we assessed that the model could be used to get a reasonable estimate of the dynamic compression forces in the lumbar spine during MMH. However, major discrepancies were also identified during these tasks, particularly with respect to the trunk forward flexion angle (rRMSEs ranging from 134 to 199%).

Generally, the IMC-PGRF model estimates of the L4-L5 JRFs did not seem to be particularly affected by the increments in load or potential inaccuracies in the predicted GRF&Ms. During symmetrical and asymmetrical lifting, the rRMSEs for the A-C (23-34%) and A-P shear forces (34-50%) were very similar across all load instances (from 5 to 20 kg). The M-L shear force was generally of very low magnitude across all the analyzed tasks, which makes comparisons difficult due to the low signal-to-noise ratio. Hence, the selected tasks did not result in sufficiently high M-L shear forces to enable an appropriate evaluation, possibly due to an insufficient degree of load asymmetry. The vertical GRFs generally showed moderate to excellent correlations (from 0.51 to 0.96) and low magnitude differences (RMSEs ranging from 5.1 to 12 %BW) between the IMC-PGRF and OMC-MGRF models. When comparing the OMC-MGRF and OMC-PGRF models (see Paper I), which uses the same kinematic input data, the correlations (from 0.86 to 0.98) and magnitude differences (RMSEs ranging from 2.3 to 8.7 %BW) showed better agreement, but the results were similar overall. These results indicate that the discrepancies in the IMC-PGRF model's estimates of spinal forces most likely stemmed from the differences in kinematic data and scaling techniques, and not the predicted GRF&Ms.

When evaluating the results for the L4-L5 JRFs and vertical GRFs against previous literature, the discrepancies identified in Paper I were generally higher [147,179,180]. However, direct comparisons are difficult in regards to the L4-L5 JRFs, as these forces have not previously been estimated based on IMC data to my knowledge. Faber et al. [179] found coefficients of determination (R^2) above 0.99 and RMSEs below 10 Nm (approximately 5%) for the L5-S1 extension moment during trunk bending when comparing an IMC-driven biomechanical model with predicted GRF&Ms to an OMC-based model. For the vertical GRFs, the R^2-values were above 0.98, while the RMSEs were below 10 N (approximately 1%). Kim and Nussbaum [180] also compared an IMC-driven biomechanical model, where the GRF&Ms were measured using pressure insoles, to a model driven by OMC and force plate data during a series of MMH tasks. They found mean absolute errors of 4.4-14, 0.16-1.4 and 1.3-2.5 Nm for the L5-S1, shoulder and knee flexion-extension moments, respectively. Overall, the peak joint moments showed relative errors of 24 (lateral bending), 38 (rotation) and 14% (flexion-extension). Finally, Karatsidis et al. [147] used a similar model and approach as in Paper I for estimating lower extremity kinetics during gait. For example, they found Pearson's correlation coefficients of 0.80-0.97 and 0.82-0.91, and rRMSEs of 7.7-19% and 14-26% for the vertical GRFs and knee JRFs, respectively.

For the trunk kinematics, our evaluation also showed larger errors overall compared with previous research [147,180]. For example, Kim and Nussbaum et al. [180] found mean absolute errors of 1.3-4.8°, 1.5-3.1° and 1.4-4.0° for L5-S1 flexion-extension, rotation and lateral bending during MMH, respectively. For the symmetrical lift from the ground in this study, the mean and peak absolute errors for the L5-S1 flexion-extension angle were 4.8° and 7.2°, respectively, which is considerably lower than the results of Paper I. Karatsidis et al. [147] found that the IMC-PGRF model showed similar trends in the lower extremity joint angles compared with the OMC-based model with Pearson correlations of 0.95, 0.99 and 0.99, and rRMSEs of 13, 7 and 14% for ankle plantarflexion, knee and hip flexion, respectively.

It is important to note that the study of Faber et al. [179] and Kim and Nussbaum [180] used different biomechanical models for their analysis, while the study of Karatsidis et al. [147] analyzed gait at different speeds, which may explain some of the discrepancies between studies. In addition, it should also be noted that the OMC systems used as the reference data for the comparative studies are also associated with errors, especially due to soft-tissue artefacts [181]. However, there could be several reasons for the large differences in trunk kinematics between the IMC-PGRF and OMC-MGRF models observed in Paper I. For example, inaccurate placement of the pelvis IMU may have inhibited the system's ability to measure the relative angle between the trunk and pelvis. As most of the subjects were shirtless during the experiment, the IMUs were taped to the skin and not attached with the accompanying velcro straps. For this reason, the pelvis IMU was placed too superiorly on the subjects' back to avoid placing it by their waistband, which did not abide with the recommended protocol, in which the pelvis IMU has to be placed on top of the sacrum. In addition, magnetic distortions may have caused inaccuracies in the IMU's orientation and position estimates over the course of the measurements, as the subjects were standing on force plates. Although not reported in Paper I, there seemed to be considerable inaccuracies in the ankle angles for the IMC-PGRF model, which further caused inaccuracies for the knee angles. This was particularly apparent for the subjects that used a squat-lifting technique, which require a high degree of ankle dorsiflexion and knee flexion. The IMC-PGRF model did not accurately replicate these movements, which resulted in the models standing more on the toes with excessive knee flexion as well as a more upright trunk posture compared with the OMC-MGRF model. This in turn, could have contributed to the underestimation of the trunk forward flexion angle.

It should be noted that the IMC-based kinematic data presented in Paper II was extracted directly from the Xsens software and not first input to the AMS as in Paper

I and III, which uses another kinematic model with slightly different joint definitions. The implementation of the IMC data into the AMS may itself cause inaccuracies, which should be considered when viewing the data. Furthermore, updates to the Xsens MVN firmware and software seemed to improve the quality of the kinematic data from Paper I to Paper II and III, most notably for the lower extremities and trunk. This was probably partly due to the pelvis IMU being positioned more correctly, but differences in lifting techniques, calibration procedures and magnetic interference between studies may also have played a role in these potential improvements. Although speculative and mostly based on visual inspections of the models, the lower extremity and trunk kinematics generally appeared more accurate and consistent in the dataset related to Paper II and III. For example, the much larger trunk forward flexion angles (Paper II) and A-P shear forces (Paper III) found in these studies indicate that the underestimation of these variables was less severe. However, future studies should further analyze the accuracy of the IMC system during lifting activities, particularly in combination with the kinematic model of the AMS, to ensure a sufficient accuracy of the kinematic data.

4.1.2. RISK ASSESSMENT

Based on the risk assessment in the two supermarkets, we identified several MMH tasks and other work factors that may pose a risk for developing WRMD. The recommendations made based on the use of sEMG (Paper II) and musculoskeletal models (Paper III) were generally very consistent, which supports the appropriateness of both methodologies for this type of analysis. For example, the similar trends in the muscle activities of erector spinae and trapezius descendens, and the JRFs in the lumbar spine and shoulders. Based on the results of both papers, the tasks that posed the greatest risk of developing MSDs were the handling of bananas, milk crates, cucumbers and bread. It was also shown that the starting positions and placement on the shelves had a substantial influence on the estimated joint loads and muscular efforts for these and other merchandise, meaning that there are several aspects of the handling practices that could be adjusted to reduce the musculoskeletal demands. In addition, the kinematic data presented in Paper II showed that a large proportion of the tasks involved undesirable working postures, stemming from a combination of the choice of transporting devices, shelf heights and store layout. Based on these results, interventions can be designed to target potentially hazardous tasks, the use of assistive devices and the design of the shopping area, which could reduce the risk of developing MSDs for the supermarket workers.

As mentioned above, the handling of bananas, milk, cucumbers and bread showed the highest forces and muscular efforts in the lower back. In general, the handling of bananas and milk showed the highest overall L5-S1 A-C (from 424 to 553 %BW) and A-P shear forces (from 90 to 155 %BW) as well as some of the highest erector spinae nEMG (from 40 to 61%). These tasks involved the heaviest merchandise handled on a daily basis in the two supermarkets (20.2 and 17.3 kg, respectively). Only two variations of bananas were analyzed (low and high starting position), as this merchandise was only placed on the lower shelves, but it was clear that the lower starting position increased the demands on the lower back. Six variations of milk were included, which also showed considerable differences in L5-S1 A-C and A-P shear forces as well as erector spinae nEMG between the variations in start and end position. For example, lifting milk from the low starting position (15 cm) resulted in higher musculoskeletal demands than lifting from the high position (75 cm) with differences of approximately 40 %BW in A-C force, 27 %BW in A-P shear force and 12% in erector spinae nEMG. Based on these data, it seems reasonable to recommend that the handling of bananas and milk should be minimized, and preferably only lifted from starting positions near waist height. For the heaviest tasks in the stores (e.g. bananas and milk), it might be beneficial to implement technical assistive devices to increase the initial lifting height. Lifting near waist height rather than from near the ground may significantly reduce the load on the lower back [182-184], and these devices has been shown to reduce muscular demands during stocking work in supermarkets [164].

The influence of start and end positon was also apparent for the considerably lighter tasks, cucumbers (10.2 kg) and bread (7.9 kg). When these tasks were initiated from a low to a high position, it resulted in some of the highest muscular efforts in both the lower back (from 60 to 71%) and neck and shoulder region (from 47 to 56%). These results were more or less mirrored in the JRFs, where the shoulder resultant JRFs (from 217 to 227 %BW), L5-S1 A-C (413 and 449 %BW) and A-P shear forces (114 and 126 %BW) all showed some of the highest values across the analyzed tasks. This may have been a result of high acceleration during these lifts, although this was not specifically analyzed in the study. The combined exposure to high low back and shoulder loads should warrant adjustments to the handling practices, as for instance, lifting from a higher starting position as well as not placing the merchandise on the highest shelves. This recommendation is supported by the study of Silvetti et al. [111], who showed that the muscular efforts of the shoulder and lower back increased significantly with higher shelf heights in a supermarket context. However, the horizontal locations also increased with the increased shelf height, which likely contributed to the increased muscular efforts in this study.

When viewing the results for the muscular and joint loads in the shoulders, it was clear that the heaviest merchandise lifted to the highest shelf heights required the highest efforts. This was most notable for the shoulder JRFs, which were substantially higher for bread (from 217 to 225 %BW) and cucumbers (from 220 to 227 %BW) lifted to the highest shelf heights than for the other analyzed tasks. These results were consistent with the muscle activities of trapezius descendens, where bread (from 44 to 59%) and cucumbers (from 47 to 53%) lifted to the highest shelf height also showed the highest values. However, several other tasks required comparable muscular efforts for trapezius, e.g. lifting yoghurts (from 40 to 45%) and cold cuts (from 31 to 53%) to high shelf heights, despite the lower mass of these merchandise (6.3 and 2.1 kg, respectively). When the yoghurts were placed at the far end of the highest shelf

(Yoghurts-HighToHighFar), the muscle activities (50 and 53%) were almost identical to lifting the considerably heavier boxes of cucumbers, likely due to the increased reaching distance. These results show that even merchandise of relatively low mass can require high muscular efforts in the neck and shoulder region due to other workplace factors, such as high shelf heights combined with long reaching distances.

The handling of bananas (from 705 to 848 %BW) and milk crates (from 637 to 799 %BW) also resulted in the highest overall knee resultant JRFs. Both these tasks involved walking a step with the merchandise in hand and the workers were often standing on one leg as they placed it on the shelves. For the right knee, the highest forces occurred during the handling of milk crates, typically while placing the crates on the shelves during which many of them were standing on or mostly supporting themselves with one leg. For the left knee, the highest forces occurred during the handling of bananas, either at the initiation of the lift, where the workers were squatting down, or when placing the merchandise on the shelf, where several of the workers mostly used on leg for support. In general, the mass of the merchandise was the main predictor of high knee forces, while the peak forces typically occurred as a result of either standing on or mostly supporting their weight with one leg.

From the kinematic data, several postural risk factors were identified in the two supermarkets. First, 22 of the 50 analyzed tasks involved forward flexing the trunk 50° or more. All of these tasks involved lifting from or to a low position (13.5-18.5 cm above floor level). Trunk flexion [28,36] and awkward postures [2,16-18,31] has been associated with the development of WRMD: for example, Punnet et al. [36] found odds ratios for developing back disorders of 4.9 and 5.7 for mild (20-45°) and severe trunk flexion (> 45°), respectively, in a sample of autoassembly workers. Second, 22 of the 50 tasks involved flexing the shoulders more than 90°. Two scenarios resulted in these relatively high shoulder flexion angles, which were lifting to high and low shelves. The most strenuous of these scenarios was lifting to the highest shelves, which resulted in substantially higher trapezius muscle activity compared to placing merchandise at low shelves. As lifting to above shoulder height [2,23-25], MMH [23,24] and a combination of these exposures [26] have been identified as risk factors for developing shoulder disorders, it is concerning that so many of the common tasks in supermarkets expose the workers to these risks. Finally, in addition to exposing the workers to higher muscular and joint loads, the low shelf heights also required a high degree of knee flexion. The 10 highest ranked tasks for knee flexion all involved lifting from or to a low position and resulted in knee flexion angles of 89° or more. This indicates that the workers adapted a squat-lifting technique when handling merchandise at low heights, possibly in an attempt to transfer some of the musculoskeletal load from the lower back to the knees. Squat lifting may reduce the load on the lower back in some lifting situations, particularly the A-P shear forces, but may also entail increased metabolic demands [185,186]. It was also common for the workers to perform light stocking work in a kneeling position, although this was not directly analyzed in Paper II. Lifting, squatting, kneeling or a combination hereof have all been associated with an increased risk of developing knee disorders [16,18]. In view of these results, it seems reasonable to recommend that the low shelves should be avoided whenever possible to reduce the workers' exposure to these risk factors.

4.1.3. EFFECTS OF LIFTING FACTORS

The main findings of the laboratory-based lifting analysis were that all the tested conditions had a significant overall effect on the peak loading of the involved joints with some lifting factors showing a substantial influence on one or more joint forces.

LM was the lifting factor that had the largest effect on the joint forces overall, particularly the L5-S1 A-C (from 472 to 672 %BW) and A-P shear forces (from 48 to 67 %BW). These results generally support the findings of previous research, which showed increased low back loading as a result of increased LM, both in terms of the low back moments [182,184,187] and L5-S1 JRFs [103,187,188]. Furthermore, the increments in LM also resulted in significantly higher knee (from 339 to 547 %BW on average) and shoulder resultant JRFs (from 40 to 127 %BW on average). Lavender et al. [125] similarly found an increased load in the knees with increments in LM during symmetrical lifting, while Schipplein et al. [122] found a reduction in the dynamic peak flexion-extension moment in the right knee (from 53 to 13 Nm) with increments in LM. This reduction was attributed to their subjects adapting a stoop-lifting technique when lifting heavier loads, which was not seen in the present study. Finally, Faber et al. [116] found that increments in LM significantly increased the load on the glenohumeral joint during MMH, which is in line with the results of Paper IV.

Increments in the AA had a negligible effect on the L5-S1 A-C forces (from 513 to 543 %BW), but a substantial effect on the L5-S1 A-P (from 52 to 68 %BW) and M-L shear forces (from 14 to 40 %BW). These results partially support the findings of previous research [101,102]. For example, Gallagher et al. [102] found no significant change in peak L5-S1 A-C force between symmetrical and asymmetrical lifting in stooped and kneeling postures based on quasi-static analysis, but significant increases in the peak A-P (~55 N) and M-L shear forces (~4.7 N). Marras and Davis [101], however, found a 5% increase in A-C force, 8% decrease in A-P shear force and 58% increase in M-L shear force when the load was placed with a 60° AA to the right of their subjects compared to a symmetrical lift. In Paper IV, the same forces increased with 2.4, 40 and 419% when lifting with a 60° AA compared with the otherwise identical, but symmetrical lift, VL-30. In addition to the increased spinal forces, the increments in AA also led to increased JRFs in the knees and shoulders. Specifically, the right knee resultant JRFs increased significantly with increments in the AA (from 439 to 679 %BW), while the resultant JRFs in both shoulders likewise increased (from 60 to 110 %BW on average). Hence, asymmetrical lifting should be avoided whenever possible, as it significantly increases the peak shear forces in the lumbar spine as well as the resultant JRFs in the knees and shoulders.

The increments in HL did not seem to have a substantial effect on the L5-S1 JRFs. More specifically, the L5-S1 A-C and A-P shear forces increased from 502 to 549 and 50 to 55 %BW, respectively, when the HL was increased from 30 to 60 cm. The changes in the M-L shear force were negligible. Previous modelling studies have shown increased L5-S1 flexion-extension moments with increased HL [123,125,189]. These studies generally showed larger relative increases than in Paper IV. However, other studies did not show this effect [190,191], possibly due to the fact that their subjects were more free to adapt their lifting technique in response to the increased HLs [190]. Schipplein et al. [123] is one of the few previous studies that has analyzed the effect of HL on knee joint moments. They found that the knee flexion-extension moment at the time of peak L5-S1 moment shifted from extension to flexion between HLs of 20 and 40 cm, and remained constant when the HL further increased up to 60 cm. The present study found that both the knee (from 375 to 484 %BW on average) and shoulder resultant JRFs (from 53 to 103 %BW on average) increased significantly with increments in HL. These findings suggests that the knees and shoulders may be the joints most affected by increments in HL.

Finally, increased VLs did not lead to increased L5-S1 JRFs, but this was mostly due to the fact that the peak forces occurred during the initiation of the lifts rather than when the box was placed on the shelves. From the time-series curves of the JRFs (see Paper IV), it seems that the L5-S1 A-C and A-P shear forces during the deposit phase were reduced as a result of increments in the VL, meaning that a lower deposit height led to higher forces in the lumbar spine. However, the increments in VL led to a substantial increase of the peak resultant JRFs in the shoulders (from 115 to 243 %BW on average), which is consistent with previous studies analyzing the effect of VL on the shoulder flexion-extension [192] and total moment [116]. Hence, there seems to be a trade-off between loading the shoulders or the lumbar spine depending on whether loads are placed at high or low positions.

4.1.4. BIOMECHANICAL TOLERANCE LIMITS

As described in Paper III, we chose to compare the L5-S1 A-C and A-P shear forces to the well-known tolerance limits of 3400 [54,88] and 1000 N [84,89], respectively. There were 8 and 6 tasks where the upper confidence limit exceeded these tolerance limits, as for instance, Bananas-LowToLow (4188 N for A-C and 1191 N for A-P shear), Bananas-HighToLow (4088 and 1097 N), Milk-LowToLow (3854 to 1113 N), Milk-LowToMid (3811 and 1069 N) and Milk-LowToHigh (3775 and 1062 N). These findings were central for recommending that the handling of bananas and milk crates should be reconsidered in the participating supermarket company. The reason for using the upper confidence limit was twofold: first, it was the more protective approach, as the upper confidence limit express the uncertainty in the force estimates, meaning that it would be statistically unlikely that the forces actually exceeded this limit. Second, the L5-S1 A-C and A-P shear forces were most likely underestimated in the field study, as described above.

The appropriateness of the compression tolerance limit is questionable [91] and it should be used with caution. It is highly generalized, as it do not control for age, sex and body weight, and largely based on expert consensus. Furthermore, the tolerance limit for A-P shear is mostly based on mechanical testing of the shear tolerance of cadaveric spinal motion segments [84,89], which do not accurately replicate real in vivo conditions. Another issue is that the lifting studies that formed the basis for the tolerance limits for A-C used 2-D static biomechanical models, which considerably underestimates these forces compared with dynamic models [95-98]. Hence, 3-D dynamic biomechanical models would certainly have produced higher force estimates for the same lifting tasks in these foundational studies (e.g. Chaffin et al. [51]), which may have led to different conclusions. However, the suggested tolerance limits may still be the most well-founded criteria that exist in the biomechanical literature today, and provide a reasonable starting point from which to evaluate what spinal loads that can be tolerated for most healthy workers.

The complications related to the use of the compression limit of 3400 N can be exemplified by reviewing the data from the laboratory-based lifting analysis (Paper IV). For example, when lifting loads between 5 and 25 kg from a pallet with an initial lifting height of 25 cm (hands to floor) and a HL of 45 cm, the L5-S1 A-C forces ranged from 3734 to 5315 N, all of which exceeded the action limit suggested by NIOSH. Actually, all of the lifting conditions analyzed in Paper IV exceeded the NIOSH action limit, which mostly involved lifting a 10 kg box to various locations. Interestingly though, none of the lifts exceeded the tolerance limit for A-P shear (from 352 to 537 N). So would it be reasonable to recommend that all these lifts should be avoided? Probably not. However, it may be a good indication that the circumstances of the lifts could be improved. In the case of handling bananas and milk crates in supermarkets, these tasks did not only show the highest A-C forces, but also some of the highest measured muscular efforts in the lower back as well as the highest forces in the knees. Hence, this may be a good place to start if one wishes to reduce the risk of developing MSDs during stocking work in supermarkets. It may also be worth asking the question: is it strictly necessary to expose the workers to these risks in order to achieve a satisfactory economic output? Probably not.

4.1.5. FIELD VERSUS LABORATORY-BASED ANALYSIS

Some interesting differences were found when comparing the results of the field and laboratory-based analyses. First, the L5-S1 A-C forces were generally larger in the laboratory study, while the A-P forces were considerably lower. In the laboratory setting, the A-C and A-P shear forces ranged from 471 to 671 %BW and 45 to 68 %BW, while the corresponding results from the field-based analysis ranged from 348 to 553 %BW and 67 to 155 %BW. The most striking difference was for the A-P shear forces, as the highest measured shear force in the laboratory study corresponded to the lowest value found in the field. This is despite these forces potentially being underestimated in the field study. The primary reason for this discrepancy may be attributed to differences in lifting techniques. In the laboratory study, the subjects mostly adapted a squat-lifting technique with a high degree of anterior pelvic tilt and were not able to move their feet away from the force plates. In the two supermarkets, there were much more variability in handling practices. The workers adopted a mixture of stoop and squat lifting, and often used a split stance with one leg in front of the other when lifting or depositing the merchandise. This resulted in a higher degree of trunk flexion and hence, a higher L5-S1 A-P shear force. These differences in lifting technique also resulted in substantial differences in the knee resultant JRFs, which ranged from 258 to 679 %BW in the laboratory study and from 495 to 848 %BW in the field. Again, this was mostly a result of the supermarket workers adopting a split stance during lifting, where they supported most of their weight on one leg. There were generally good agreement between studies for the shoulder resultant JRFs. For example, lifting cucumbers (10.2 kg) to the highest shelf heights (108 and 141 cm above floor level) resulted in shoulder JRFs of 226 %BW on average, while lifting the 10 kg box to the highest shelf height (150 cm) in the laboratory setting resulted in JRFs of 243 %BW. Similar trends were found for the lower shelf heights.

These discrepancies between studies may be indicative of the limitations associated with analyzing work-related lifting in a laboratory setting. For example, restricting the foot placement prevents the subjects from adapting the most efficient technique in response to the mass and position of the load. Furthermore, there might be a tendency for the subjects to lift in a more controlled fashion, potentially adapting the lifting technique that they think will be most safe, e.g. the squat-lifting technique seen in the laboratory study. However, when handling materials at work, subjects may adapt other handling practices, which may be mostly based on overall efficiency in terms of work output, rather than safety concerns. Hence, field-based analysis will more likely be representative of the actual handling practices. However, laboratory studies gives the researcher more control of the experiment and enables a more detailed evaluation of individual lifting factors in isolation. This in turn, will be more useful when providing general recommendations for the regulation of MMH across multiple industries.

4.2. PRACTICAL IMPLICATIONS

The results of the dissertation have practical relevance in multiple areas, specifically in relation to the use of the presented methodology for risk assessment of MMH, the handling practices in the supermarket sector as well as improving the basis for advising on the regulation of MMH across multiple industries.

4.2.1. APPLICABILITY OF THE METHODOLOGY

The methodology for risk assessment of MMH based on IMC and musculoskeletal models provides some exciting new opportunities. It enables the simultaneous estimation of multiple joint loads during MMH in the field. As can be seen from Paper

III and IV, certain lifts and lifting factors substantially influence multiple joint forces, and may produce critical loading of one joint, but not another. Hence, to ensure that a risk assessment leads to recommendations that protect the workers against most hazardous loads, one needs to be able to estimate the loads in multiple joints accurately. As the AMS model is highly detailed and mostly based on cadaver studies, it may be one of the methods with the most potential for estimating accurate in vivo loads in multiple joints during MMH. As the implemented technologies will undoubtedly improve over the coming years, the limitations identified in the present study may be overcome, which could greatly enhance the applicability of the methodology. However, there is still some way to go before the methodology can be used as a standard ergonomic tool for risk assessment of MMH. It is very time-consuming, require highly specialized skills and is mostly applicable for in-depth biomechanical analysis of two-handed lifts with decent hand coupling, meaning that the hand placement will more likely be consistent between trials.

4.2.2. HANDLING PRACTICES IN THE SUPERMARKET SECTOR

Based on the risk assessment studies, several aspects of the handling practices in supermarkets should be reconsidered to reduce the risk of developing MSDs for the workers. This may include avoiding specific MMH tasks, re-designing shelves and the store layout as well as implementing technical assistive devices. First, the handling of bananas and milk crates should optimally be avoided, while bread and cucumbers should not be lifted to the highest shelf heights. Furthermore, it would be advisable to avoid lifting the heavier merchandise from or to low positions, while light tasks that require highly awkward postures, such as ColdCuts-HighToHighFar, should have more accessible placements on the supermarket shelves. Second, the large proportion of the analyzed tasks that required undesirable working postures could be minimized by re-designing the shelves. For example, the lowest and highest shelves could be removed, while the shelf depth could be reduced, which could potentially minimize hazardous working postures. This would also mean that fewer merchandise could be stocked in the shopping area, which may affect sales. However, the participating supermarket company was in the process of re-designing many of their stores during the project period, where these adjustments were actually implemented to a large extent. Overall, they have created more spacious stores with less merchandise in the shopping area, meaning that the shelves were not as low or high as before, and the aisles were wider, making it easier to transport pallets or assistive devices in to the shopping area. Whether these changes were motivated by work environment issues is unclear, but it shows that it is definitely possible to address design issues that may lead to improvements in the physical work environment. Finally, the use of technical assistive devices should be encouraged, which can be facilitated by creating more aisle space in the shopping area. This would help avoid lifting from low positions, whether full boxes are lifted or the merchandise is stocked individually. As the MMH tasks analyzed in the dissertation are common throughout the supermarket sector in Denmark, these practical recommendations apply to the industry as a whole.

4.2.3. REGULATION OF MANUAL MATERIAL HANDLING

Regulating MMH across industries is complicated, as many factors affect the associated health risk, making it difficult to provide clear and appropriate limits for handling operations. If the limits are too general and vague, they may not be sufficiently protective for the workers. If they are too specific and strict, they may be hard to follow and entail great economic disadvantage for the industries most affected. However, a great body of literature exists on the health risk of performing work-related MMH, which is not currently used to its full potential. In Denmark, the handling guidelines published by the Danish Working Environment Authority are grossly simplified and the inspection of work sites largely relies on the knowledge and experience of the supervisor [193]. It can not be expected that each of these individuals will possess adequate knowledge of the biomechanical consequences of performing MMH to provide meaningful and consistent recommendations. This dissertation as well as many other previous studies, provides a great foundation for creating more industry specific and scientifically substantiated recommendations for work-related MMH. For example, in the supermarket sector, weight limits for specific handling situations can be formulated, which may be specified in terms of starting positions, shelf heights, load asymmetry, frequency and duration among other things. Musculoskeletal models may be useful in this regard, as the loads in multiple joints can be analyzed for a wide range of handling situations over the complete lifting cycles, hereby facilitating the identification of specific time points where loads may become critical. In addition, the models provide a great tool to visualize the lifting situations that may be hazardous, which would greatly benefit the individuals having to make the decisions about the health risk of MMH at various workplaces. Future research should review and compile the most relevant information from the scientific literature on the health risk of MMH to provide more substantiated and useful recommendations, similar to the 1981 guidelines of NIOSH [54].

4.3. LIMITATIONS

The dissertation had several limitations, which were described in detail in the appended papers. The most important of these limitations are summarized in the following.

The main limitations associated with the evaluated methodology (Paper I) was the underestimation of the trunk flexion angle and A-P shear force. As was discussed in the preceding sections, inaccuracies in the placement of the pelvis IMU as well as potential magnetic interferences influencing the estimation of the lower extremity kinematics were likely contributing factors. Furthermore, the study only evaluated the accuracy of the trunk kinematics, L4-L5 JRFs and vertical GRFs, while the proceeding studies estimated the L5-S1, knee and shoulder JRFs. Hence, the accuracy of these force estimates were not documented for the IMC-PGRF model prior to performing the risk assessment. This is also true for the hand forces, which were predicted using contact elements. Optimally, the hand forces should have been measured using force transducers to evaluate the accuracy of the model predictions, which would be relevant to evaluate in future studies. Paper I, III and IV were all affected by this limitation. The differences in determining the box kinematics between the two evaluated methodologies (IMC-PGRF vs. OMC-MGRF) may have introduced discrepancies not related to the models themselves. Additional IMUs could have been implemented to measure the box kinematics for the IMC-PGRF model, which would have minimized the potential errors stemming from this issue. This limitation is associated with both Paper I and III. Finally, it is important to recognize that the model based on OMC and force plate data is not a true gold standard, as there are also well-known inaccuracies related to the use of marker-based motion analysis, such as soft-tissue artefacts [181].

The main limitations related to the risk assessment studies (Paper II and III) were related to errors in the sEMG and kinematic measurements, where the latter issues additionally caused errors in the musculoskeletal model analysis. These errors led to the exclusion of a large proportion of the analyzed tasks. For the sEMG-measurements, signal dropout and poor skin-electrode contact were the predominant issues, which is a well-known limitation of sEMG [194,195]. For the IMC data, errors in the system's estimation of the hand positions led to large errors in the orientation of the smaller, narrow boxes in the musculoskeletal model analysis. Furthermore, the kinematic data also showed excessive palmar dorsiflexion, particularly during one-handed handling, which led to errors in the muscle wrapping of the wrist flexors. Based on these limitations, it was concluded that the methodology was best used for in-depth analysis of the relatively heavy merchandise in larger boxes, which could potentially be identified as hazardous using observational methods. Another limitation of these studies was that we had to impose lifting techniques on the workers to some extent to facilitate the musculoskeletal modelling procedures. Hence, we did not accurately capture the natural intra and inter-subject variability associated with handling materials in supermarkets. Finally, magnetic distortions may have caused orientation and positional drift in the kinematic data, which is a well-known issue related to the use of IMC. To what extent drift was present in the measurements is unclear, but we were well aware of this issue doing data collection and performed frequent calibrations to minimize its influence. In addition, significant efforts have been made by the developers of the IMC system to correct for drift using an advanced Kalman filter [146] as well as a post-processing tool (HD-reprocess) in the accompanying software.

Besides what has already been mentioned, the main limitations associated with Paper IV were related to the restrictions imposed by the measurement equipment, which inhibited the subjects' ability to adapt their technique in response to changes in load mass and position. This limitation exemplifies the importance of the developed methodology for field-analysis, as the use of GRF prediction and the on-body IMC system allows the subjects to more freely adapt their lifting technique.

Finally, the use of peak values for both the muscle activities, joint angles and JRFs does not take full advantage of the information contained in the dataset. This is a limitation of all the included studies. Implementing measures of musculoskeletal load over the complete lifting cycles (e.g. integrated sEMG or joint impulse) as well as cumulative workload could have provided valuable information to both the field and laboratory-based analysis. Furthermore, it would have been appropriate to incorporate statistical methods that take advantage of the whole time-series curves of the sEMG, kinematic and force data, such as statistical parametric mapping [196].

4.4. CONCLUSIONS

In this dissertation, a novel methodology for estimating dynamic joint loads during MMH in the field was developed and evaluated based on musculoskeletal models driven by IMC data and predicted GRF&Ms. The methodology was used in combination with sEMG to conduct a risk assessment of the stocking work in two supermarkets. Similar models were also used to evaluate the relative importance of several lifting factors on multiple joint loads in a laboratory setting.

The evaluation study showed that the IMC-PGRF model was able to estimate dynamic compression forces in the lumbar spine with a reasonable accuracy compared with a model driven by OMC and force plate data. However, discrepancies between the models were also identified, particularly for the trunk flexion angle.

In the risk assessment studies, several MMH tasks that may pose a risk for developing WRMDs were identified, particularly the handling of bananas, milk crates, cucumbers and bread. This inference was made based on musculoskeletal model estimates of the dynamic L5-S1 A-C and A-P shear forces, which exceeded well-known tolerance limits suggested in previous literature, as well as the tasks requiring relatively high muscular efforts in the lower back and shoulder and neck region. Furthermore, a large proportion of the analyzed tasks involved undesirable working postures, e.g. a high degree of forward bending, lifting to above shoulder height and excessive squatting during lifts. Based on these findings, we recommended that several aspects of the handling practices in the participating supermarket company should be reconsidered. This could involve re-designing shelves, increasing the aisle space in the shopping area, implementing technical assistive devices to increase the handling height as well as reconsidering the placement of certain merchandise on the shelves. However, due to the high musculoskeletal load associated with the handling of bananas and milk crates, individual stocking should be considered, but limiting the size of the boxes as well as the amount of merchandise they contain may be warranted.

The laboratory-based lifting analysis showed that LM had the largest effect on the peak joint forces overall, particularly the L5-S1 A-C and A-P shear forces, while the AA had a negligible effect on the A-C force, but a substantial influence on the L5-S1 shear forces as well as the knee and shoulder resultant JRFs. Increments in the HL did not substantially influence the forces in the lumbar spine, but significantly increased both the knee and shoulder resultant JRFs, while the VL had the most substantial effect on the shoulder resultant JRFs. Based on these results, it appears that LM is the most important factor to consider for reducing joint loads overall, while asymmetrical lifting should be avoided, as it substantially increases the shear forces in the lumbar spine as well as the peak forces in the knees and shoulders. The HL seems less important in regards to the spinal loads, but exhibit a greater influence on the forces in the knees and shoulders. For the VL, there appears to be a trade-off between loading the lower back or shoulders depending on whether the load is placed on low or high shelves. Hence, peak joint loads could be reduced by limiting the amount of tasks that require lifting materials to positions close to the ground or above shoulder height.

The dissertation represents the first comprehensive analysis of MMH based on state-of-the-art musculoskeletal models, both in a field and laboratory setting, while the developed methodology provides new opportunities for in-depth biomechanical analysis of MMH at the workplace.

4.5. PERSPECTIVES

As mentioned previously, the dissertation provided a novel methodology to estimate dynamic joint forces during MMH in the field. Despite its current limitations, the methodology shows great potential for field-based risk assessment of MMH and may become a valuable tool for future studies. Musculoskeletal models have the great advantage of being able to estimates forces in multiple joints and muscles simultaneously, which is important when analyzing complex full-body tasks, such as MMH. During the project period, improvements have been made by the developers of the software (AMS) and hardware (Xsens), meaning that many of the issues we encountered during the process of evaluating and applying the methodology may already have been solved. As these technologies will undoubtedly improve further over the coming years, the accuracy and versatility of the presented methodology will likely improve as well. In view of this, I believe that the use of musculoskeletal models for biomechanical analysis in the field will become more widespread in the coming years and provide new insights in to the dynamic loading of the joints in real-life work scenarios.

One such example was provided in the dissertation, as we implemented the methodology for the first comprehensive risk assessment of MMH in the supermarket sector, which led to the identification of several tasks and work factors that may contribute to the development of WRMD. Several recommendations were made based on these results, which could potentially help improve the working conditions for supermarket workers in Denmark and elsewhere. For this to happen, I believe it is important that supermarket companies do more to prioritize their workers' health and well-being, and not let concerns about profit or efficiency stand in the way of what is often simple and relatively inexpensive work environment initiatives. The results of the risk assessment could potentially have far-reaching implications for the physical work environment in the supermarket sector, depending on the commitment of the major supermarket companies.

The dissertation also provided the first comprehensive laboratory-based lifting analysis using state-of-the-art musculoskeletal models to determine the relative importance of several lifting factors on multiple joint loads. This study exemplified the importance of assessing multiple joint loads simultaneously in order to adequately assess the physical demands of MMH. These and other previous findings in this area should be implemented to a larger extent when formulating recommendations and regulations for work-related MMH. In my view, musculoskeletal models could be a great tool to formulate more detailed recommendations and specify weight limits in relation to various load positions. Furthermore, the identification of the relative importance of lifting factors on joint loads from this dissertation may help the supervising entities identify which lifting factors to prioritize and hereby, qualify their assessments of MMH at the workplace. Finally, these models may also be a valuable tool to visualize the handling situations leading to critical joint loads.

Collectively, these studies provided novel contributions to the ongoing efforts to reduce the incidence of WRMD during MMH, and can be used to improve current handling recommendations, particularly in the supermarket sector.

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APPENDICES

Appendix A. Musculoskeletal modelling of manual material handling in the supermarket sector: full dataset

Appendix B. Papers I-IV

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